Reliability of Isometric Neck Strength and Electromyography Measures Relevant for Concussion Prevention in Athletes

by

Sivan Almosnino

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Abstract

The purpose of this investigation was to assess the between-day reliability of selected force-time curve indices and the activation onset of selected neck muscles in the performance of maximal, isometric contractions in five different directions. The measures extracted are deemed important for future investigations aimed at exploring the role of cervical musculature in reduction of concussion occurrences and severity in sports.

Twenty eight physically active male participants performed two testing sessions separated by 7-8 days. In each testing session, force and surface electromyography (EMG) data were recorded simultaneously in a custom-made testing apparatus whilst subjects performed four randomized maximal isometric efforts in extension, flexion, and left and right lateral bending and protraction. The variables examined were the peak force, rate of force development (RFD), time to 50% of peak force and bilateral activation onset of the splenius capitis, upper trapezius, and sternocleidomastoid. For all variables, reliability was assessed by: 1) difference scores between the testing sessions and corresponding 95% confidence intervals; 2) standard error of measurement (SEM), expressed in either the original units of measurement, or as a coefficient of variation; and, 3) Intraclass correlation coefficients (ICC). The results indicated that for all variables, in all testing directions, no differences in scores were observed between the first and second testing sessions. The precision of measurement for all measures, barring muscle activation onsets obtained in protraction, was deemed acceptable for future clinical application. ICC score ranges for force-time curve-based measurements were high (< 0.90), while for muscle activity onsets, the ICC ranges were low to moderate (0.23 -0.79).

Based on these results, it was concluded that, in highly active male participants, a dedicated familiarization session for the elimination of potential learning effects is not required. In addition, for the majority of testing directions, the force-time curve-based variables as well as

muscle activation onsets are recorded with a sufficiently high level of precision, which make them prime candidates for use in future investigations concerned with quantitative assessment of cervical musculature function.

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Abbreviation of Terms

Muscles:

SCM	Sternocleidomastoid
SpL	Splenius Capitis
TRP	Trapezius

Direction Efforts:

Ext	Extension
Prot	Protraction
RLB	Right Lateral Bend
Flx	Flexion
LLB	Left Lateral Bend

Dependent Variables:

RFD	Rate of Force Development
PF	Peak Force
$T_{50}PF$	Time to 50% of Peak Force

Statistics:

CI	Confidence Interval
CV	Coefficient of Variation
ICC	Intraclass Correlation Coefficient
TE	Typical Error, similar to the SEM
SEM	Standard Error of Measurement
SDD	Smallest Detectable Difference

Chapter 1
Introduction

Concussions due to sport participation have received much attention in both the popular press and the peer-reviewed literature, primarily due to the detrimental short and long-term effects on brain function [3,10,12]. The consequences of concussions may include permanent learning disabilities, impaired development of social and cognitive skills and psychiatric problems manifested in forms of depression and violence [11].

The extent of this problem can be exemplified by examination of relevant injury surveillance reports; Canadian minor hockey leagues report that 10 -12% of players sustain a concussion each year, for an estimated 64,290 hockey-related concussions detected yearly [7]. In the United States, an estimated 300,000 concussions occur annually in athletic settings with high rates of concussions documented particularly in hockey, football, rugby and soccer participants [6, 13]. Within this framework, it appears that females and younger players seem to be more susceptible to injury [17]. In addition, it is claimed that many concussion incidents are not reported or go undetected, which results in underestimation of what is already considered a worrisome number of injuries [23].

A multi-tiered approach has been used in recent years in an attempt to reduce the incidence and severity of concussion in sports, including the improvement of protective equipment [2], design and implementation of educational programs for coaches and players [19], alteration of playing rules [11,14], and reliance on comprehensive neuropsychological and motor control assessments to evaluate the severity of injury and to guide decision-making around return to play [1,11]. In addition, innovative modeling of the neck and head in conjunction with reconstruction of injury incidents in professional football has provided a better understanding of injury mechanisms [15,21,22].

An additional preventive strategy that has been advocated relates to the proper conditioning of the cervical musculature in those participating in sports [9,20]. The underlying theory is that potentially harmful impact forces delivered to the head might be attenuated if the neck muscles are activated prior to collision, as this enhances the stiffness of the head and neck complex [22]. While incorporation of cervical musculature conditioning programs as part of regular training regimens is feasible from both a time and cost perspective, it requires the ability to measure cervical musculature capabilities with an acceptable degree of reliability. This is an essential prerequisite for the design of interventional programs, as well as the assessment of their effectiveness.

The ability to reliability measure cervical strength (CS), as well as the coordination of the neck musculature using surface electromyography (EMG) is challenging due to basic methodological limitations [4,16,18]. In an attempt to address these, and supported by recent efforts to improving experimental methods [5,8], a testing apparatus capable of measuring three dimensional isometric neck muscle force was constructed at the Human Mobility Research Centre at Queen's University. As with any measurement tool, establishing the reliability of the testing apparatus is the first step in ensuring the device can be used in clinical assessment and decision-making processes.

The purpose of this investigation, therefore, is to determine the between-day reliability of a custom-made dynamometer in recording various neck muscular strength and EMG parameters in a neutral sagittal head posture while performing isometric maximal voluntary exertions (MVEs) in five different directions. This study differs from the majority of previous investigations concerned with measurement of cervical musculature capabilities primarily in terms of the outcome measures reported upon. These include: the rate of force development, time needed to reach a percentage of peak force, and the activation onset of selected neck muscles relative to the

onset of force production. These measures were chosen in order to connect the theoretical considerations of the neck musculature's role in concussion prevention to results found in the clinical setting. The results of this study will form the basis for a future prospective study aimed at determining if there is an association between certain neck musculature parameters and the likelihood of concussion occurrence, as well as serve as a testing device for assessment of outcomes of intervention programs.

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Chapter 2

Literature Review

The following review of the literature is meant for acquaintance of the reader with current measurement methods of cervical strength (CS) and neck muscle surface electromyography (EMG). This review aims at highlighting both advantages and limitations of current methods, which were taken into account in the design of the testing apparatus used in this investigation, as well as in the formulation of the experimental protocol.

2.1 Measurement Methods of Cervical Strength Capabilities

2.1.1 Manual Muscle Testing

Manual muscle testing (MMT) of cervical muscular strength (CS) is prevalent in clinical settings, most likely due to cost and time effectiveness. MMT procedures are designed for the evaluation of joint range of motion and muscle group performance acting against gravity and examiner applied manual resistance. The scoring of the performance is based on an ordinal scale ranging from 0 to 5, where zero indicates no movement of the limb when the agonist muscle groups are not directly acting against gravity and there is no visual or palpable indication of muscle contraction. A score of five indicates that the tested joint exhibits a full range of motion when moved against gravity and that maximal strength is attainable when efforts are exerted against both gravity and examiner applied resistance [3,12,13,19,22].

The use of MMT for the assessment of muscular function has been criticized primarily due to the coarseness of the measurement scale which classifies a large portion of muscular strength capabilities as a score of four [12,13,19]. Specifically, it has been shown in different muscle groups that the percentage of muscle strength needed to overcome gravity ranges between 4-20% [12,13,22]. Such efforts are given, by definition, a score of three on the MMT scale. As

noted above, a score of five is given to maximal muscular strength capabilities. Thus, the vast portion of the strength capabilities of the muscle being tested lies somewhere within the scoring range of 3 to 5, and is extremely influenced by the examiners subjective opinion. This subjectivity leads directly to an inability to detect clinically relevant changes in performance, especially if improvements are less than 25% of pre-intervention values [12,13]. In addition, the reliability of MMT results, especially among different examiners assessing the same individual, is considered to be poor [12,19]. The poor inter-examiner reliability of MMT may be partially attributed to the lack of standardized testing procedures and also to differences in examiner experience.

2.1.2 Fixed Frame Dynamometry

The vast majority of investigations concerned with measurement of CS have used self-manufactured devices called fixed frame dynamometers (FFDs) to achieve their goal. In essence, subjects exert efforts against some type of load cell which is affixed to a stable structure, such as a wall or a rigid frame. The position of the subject relative to the load cell is often adjustable to enable accommodation of individual subject anthropometrics. The coupling of the subject's head to the device is achieved via an assortment of specialized attachments, while extraneous efforts of other muscle groups are minimized by firmly securing the trunk and sometimes the upper limbs with harnesses, seatbelts and straps. The tests are most often performed in a sitting, although testing in supine, prone and standing position have also been reported [4,7,35,55]. It has been argued that testing in a seated position is more functional with respect to the role of the neck musculature in maintaining upright posture of the head, and that better stabilization of the upper body is attainable [55]. However, a concern with seated testing is whether the subject's legs are

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being used to contribute to measured efforts, especially if the feet are placed on the ground during testing.

The most commonly described head position inside the device is the so-called 'neutral' position [15]. This position is often described as one that is self-selected by the subject according to individual comfort. Some investigations have standardized neutral position across subjects by defining the location of head's anatomical land-marks relative to the horizontal plane [27,28,60]. Several investigation report CS values for head positions other than neutral, with the majority manipulating the position of the head within ± 30° of head flexion or extension in the sagittal plane [33,46]. Others have reported CS measurements in different lateral bending head positions [45] and in different head rotation positions [65].

Measurement of CS using fixed frame dynamometry allow for the quantification of only sub-maximal or maximal *isometric* exertions. In this respect, the majority of investigations have quantified isometric efforts in directions performed along the primary movement axes, namely: flexion, extension, left and right lateral bending and, to a lesser extent, bilateral rotations. However, it is often problematic to interpret the results of different studies because of the semantics used to describe the movement direction in relationship to the placement of the support against which efforts are performed. Specifically, in almost all investigations, flexion efforts are performed against supporting pads or surfaces placed across the subject's forehead. The location of the support results in subject efforts that are more reminiscent of head protraction, or of combined head and neck flexion. In order to quantify the rotational nature of pure flexion efforts, support should be provided under the mandible, and the distinction between the directions of effort with regards to the external support should be acknowledged [15]. The same argument can be made for the movement described as extension, where the head support is located on the posterior aspect of the parietal bone. Movement efforts performed against such a support would

be suggestive of head retraction. For quantification of true extension efforts, external support should be provided under the external occipital protuberance. Developing a testing apparatus that allows measurement of true extension efforts may be challenging from a design perspective, since the shallow concavity of this anatomical region may lead to sliding of the head across the supporting surface during performance of maximal efforts.

2.1.3 Measurement of Cervical Strength using Isokinetic Dynamometers

Probably due to the technical and financial challenges associated with development and fabrication of custom FFDs, several researchers are have opted to assess CS using existing muscle strength testing devices, namely isokinetic dynamometers. Currently, manufacturers of isokinetic dynamometers do not supply specialized attachments for the head. As such, researchers have either used existing attachments (such as the one provided for the knee) [6,40], or have manufactured self made ones for coupling of the head to the testing apparatus [44,45,52]. There are several advantages to using isokinetic devices for CS testing. Most importantly perhaps is the commonality of testing platforms across different laboratories which allows for the collection and comparison of CS indices of normal and pathological populations. In addition, isokinetic devices allow measurement of dynamic CS function while at the same time enabling the examiner to have good control of subject range of motion and the type of effort performed [12]. This is especially pertinent if subjects are known to have or are claiming to be suffering from injury.

On the other hand, there are several methodological limitations associated with measurement of CS using isokinetic dynamometry. The first relates to the fact that isokinetic dynamometers measure the external force that is applied by the subject perpendicular to the lever arm [12]. As such, any efforts directed along the long axis of the attachment is not recorded because these efforts do not produce an external moment. Thus, in some cases, it would seem that

recordings might underestimate the true CS capabilities of the subject [6,52]. Another problem associated with isokinetic testing of CS is the difficulty in aligning the biological center of rotation with the mechanical axes of the testing device. In isokinetic testing of other joints, it has been found that misalignment may significantly alter the results obtained [12]. Accordingly, rigorous procedures should be employed to standardize the method of axes alignment.

2.1.4 Isometric Cervical Strength Values

Table 2.1 summarizes the results of investigations reporting peak isometric CS values performed along the primary axes of movement. The diversity in the characteristics of the testing devices, as well as differences in testing procedures have yielded a range of CS scores that are not directly comparable even for subjects matched by age, sex, anthropometrics, and physical activity levels. Consequently, the following discussion is aimed at extracting general observations regarding human CS capabilities as well as possible reasons to observed discrepancies in CS test scores across studies.

Table 2.1: Cervical strength values reported for isometric efforts in a neutral head posture*

Author(s)	n	Age	Status	Device	Position	Ext	Flex	LLB	RLB	LRot	RRot
Barton & Hayes [4]	3 ♂, 7 ♀	27.4 (7.9)	Healthy, no neck pain	Own	Supine		45 (18)				
	3 ♂, 7 ♀	42.5 (12.2)	Unilateral neck pain and head ache > 3 months				22 (13)				
Cagnie et al [7] (in Nm)	48 ♂	20-59	No neck injury 1 yr prior, no neck strength training	Biodex Isok	Supine	36.4(7.7)	24.0(6.0)				
	48 ♀	20-59				26.5(6.2)	16.6(6.6)				
	30♀	NA	neck pain			22.3(5.6)	16.7(3.3)				
Levoska et al [35]	34 ♂ Mil. recruits	18-20	5 phys activity sessions/week, no neck strength	Own	Supine	245 (42)	209 (69)	192 (34)	176 (40)		

Author(s)	n	Age	Status	Device	Position	Ext	Flex	LLB	RLB	LRot	RRot
Amudsen [1]	17 ♂ 16 ♀	NA NA	No history of neck injury	Own	Sitting	194(122) 151 (26)	133 (51) 81 (15)	139 (30)	129 (24)		
Chiu et al [8]	17 ♂ 17 ♀	20-39 19-39	"Normative active"	MCRU	Sitting	66 (26) 54 (19)	72 (25) 47 (17)				
Choi et al [10] (in Nm)	10 ♂	31 (2)	No history of neck injury	Own	Sitting	28 (3)	18(3)	16.9(2.8)	17 (2.9)		
Garces et al [21] (in Nm)	51 ♂ 43 ♂	20 to 60 +	Healthy 1 yr prior, no neck strength training	Kin Com Isok	Sitting	253 (67) 139 (51)	211 (56) 116 (45)				
Gera [22]	25 ♂	31.6 (6.2)	Healthy	Own	Sitting	146 (42)	205(62)	142 (48)	134 (49)		
Jordan et al [29] (in Nm)	50 ♂ 50 ♀	20 - 70	No neck pain 1 yr prior	Own	Sitting	55 50	30 22				
Kumar et al [32]	21 ♂ 19 ♀	24.4 (2.4) 23.9 (2.8)	No history of neck injury	Own	Sitting	100(28) 72(20)	72(18) 41(14)	76(23) 54(16)	76 (26) 52 (17)		
Legget et al [35] (in Nm)	67 ♂ 30 ♀	18-62	No history of neck injury	Own	Sitting	38 22					
Portero et al [44] (in Nm)†	77 ♂	27.6 (3.9)	Healthy	Cybex Isok	Sitting			34.1 (3.7)			
Rezasoltani et al [49]	6♂ Hockey players	18-30	Junior national team members, 6 phys activity session/week	Own	Sitting	332 (30)					
Seng et al [52]	17 ♂	24 - 38	Healthy	Biodex Isok	Sitting	45 (11)	23 (5)				
Suryanarayana & Kumar [57]	19 ♂ 20 ♀	22.3 (3.7) 22.6 (3.9)	Healthy, non active	Own	Sitting	45.1(24.3) 39.5 (25)	31.4(9.9) 19.8(10)				
Strimpakos et al [55]	17♂ 16 ♀	23.5 (5.3) 28 (11.9)	No history of neck pain	Own	Sitting/ Stand	302(63)/ 273(59) 178 (36)/ 160 (32)	229(50)/ 213(55) 101(24)/ 91(21)	241(46)/ 218(46) 122.(23)/ 116(28)	233(45)/ 218(44) 122(26)/ 117(26)	15(4.5)/ 14.7(4) 6.6(2)/ 6.8(1.9)	15(4)/ 14.5(5) 6.7(2)/ 6.6(1.7)
Valkeinen et al [59]	29 ♂	35.1 (10.7)	Healthy, non active	Own	Sitting	278 (50)	151 (47)				
Ylinen et al [64]	33 (Not detailed)	18-35	No history of neck pain. Physically inactive	Own	Sitting	195 (6)	94 (42)			8.6 (3.4)	8.4 (3.2)

Author(s)	n	Age	Status	Device	Position	Ext	Flex	LLB	RLB	LRot	RRot
Ylinen et al [66]	21 ♀	44(8)	No neck pain 6 mo. Prior	Own	Sitting	187 (39)	76 (24)			7.4 (2.3)	8.0 (2.4)
	21♀ neck pain	44 (6)	neck pain			132 (39)	54(18)			6.1 (1.6)	5.8 (1.2)
Vasavada et al [60](in Nm) £	11 ♂	32 (6)	No history of neck disorders	Own	Sitting	52 (11)	30 (5)	36 (8)		15 (4)	
	5 ♀	30 (6)				21 (12)	15 (4)	17 (8)	15 (8)	6 (3)	

♂ = males, ♀ = females. **Ext** = extension, **Flex** = flexion, **LLB** = left lateral bending, **RLB** = right lateral bending, **LRot** = left rotation, **RRot** = right rotation. All values reported as mean (standard deviation). Values were rounded to the nearest integer in certain instances.
*Strength values in N unless otherwise noted.
‡ Device: Own refers to self-fabricated testing apparatus. Isok refers to utilization of a commercial isokinetic dynamometer. MCRU is a commercialy availabe testing apparatus for cervical strength.
† Portero et al [44] combine left and right lateral bending scores.
£Vasavada et al [60] combine scores for left and right lateral bend and left and right rotation scores for males, and left and right rotation scores for females. Moments resolved at C7-T1 reported.

2.1.4.1 Influence of gender on CS scores:

When comparing non-normalized peak strength scores attained by females to those of their age- and physical activity-matched male counterparts, it is apparent that females are 20-50% weaker than males across all movement directions [4,7,8,21,31,34,56]. When expressed as peak force ratios (i.e. extension/flexion peak force or moment), Cagnie et al. [7] report that these are comparable across sexes (1.59 for males, 1.56 for females). The differences in absolute CS between males and females may be result of the smaller dimensions of females' heads and necks, although more investigations are required to ascertain this point. It should be noted that neck musculature weakness of females vs. males has been implicated as being a major contributor to females' susceptibility to neck and head injuries in sport settings [36,60].

2.1.4.2 Influence of age on CS scores:

There is some debate regarding the influence of age on CS scores; Dvir and Prusahnsky [15] summarize that a decline is observed in CS scores as a function of age, although the decline is not as pronounced as that of other muscle groups. On the other hand, Cagnie et al. [7] report

that no differences are observed in CS as a function of age. Although this latter investigation used a relatively small sample size in each age group (n=24), their results are in agreement with Salo et al. [51], who found no differences in CS peak force in 220 women divided into four different age groups. Interestingly, while Valkeinen et al. [59] report no differences in CS peak force as a function of age, they do observe differences between older and younger groups of participants in rate of force development values recorded for isometric extension efforts. Some researchers have argued that the ability of the cervical musculature to maintain their strength capabilities through the life span may be due to fact that these muscles are consistently used for head stabilization during daily activities [7,59].

With regards to adolescents under the age of 18, this group usually exhibit lower CS scores than adults, and it has been suggested that this might partly explain the predisposition of this age group to head and neck injury in sport settings [32,60].

2.1.4.3 Influence of movement direction on CS scores:

Maximal CS scores are recorded in extension, irrespective of gender or age. Flexion, protraction and lateral bending efforts CS scores usually exhibit differences of 5-15%, with a tendency for flexion/protraction to be the weaker direction. Rotation efforts produce the lowest CS scores. In healthy populations, symmetry in CS scores is usually observed between left and right lateral bending and bilateral rotation. Interestingly, there is some indication that the peak force attainable in neutral head posture is not necessarily in pure extension efforts. Specifically, Gabriel et al. [20] showed that, in subjects tested during maximal isometric exertions in multiple directions, the direction that elicited the greatest force value was 30 degrees right of extension. The authors argue that this might be due to direction dominance, as their participants were all right-handed.

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2.1.4.4 Influence of testing position on CS scores:

Testing performed in sitting seems to produce higher CS scores than supine, prone or standing position [55]. Only two studies have actually compared the influence of measurement positions using the same subjects and instruments: Strimpakos et al. [56] found that sitting produced higher peak strength than standing across all movement directions; and, Gogia et al. [23] found that prone position produces higher extension values than sitting (which does not follow the general observation stated above). Interestingly, in both studies, the between-day reliability proved to be better for the non-sitting position.

2.1.4.5 Expression of strength values with reference to the mechanical axis:

Studies that have expressed strength values in Nm inherently incorporate a length component as part of the calculation. An example of the discrepancies this might produce between studies is given by Dvir and Prushansky [15]. The authors consider the results of Olivier et al. [42], where dynamic CS values were obtained for elite rugby players using isokinetic dynamometry at 30 degrees per second, with the C7-T1 aligned with the mechanical axis of the dynamometer. For extension and flexion, the mean peak moment values obtained for 183 rugby players were 56 Nm and 39 Nm for extension and flexion, respectively. The authors then present the results of Jordan et al. [29] for age-matched, non-physically active males tested using FFD in isometric efforts where moments were 65 Nm and 37 Nm for extension and flexion, respectively. It may be argued that the comparison made by Dvir and Prushansky [15] is invalid, as the rugby players performed dynamic efforts, where as in Jordan et al. [29] the efforts were isometric (and hence values are bound to be higher in accordance with the force-velocity relationship). Even so, support to the example given by Dvir and Prushansky [15] can be found in a study by Rezasolatni et al. [50], who showed that systematic adjustment of lever arm length (which was a function of

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the location of the thoracic support in their study) resulted in significant differences in CS values. Based on the above, Dvir and Prushansky [15] recommend reporting of CS force values with no reference to the lever arm (i.e. N instead of Nm).

In an effort to compare results obtained in different investigations, several authors have reported peak force or moment ratios (i.e. isometric flexion/extension or lateral bending/extension peak strength). The use of such ratios for clinical evaluation of antagonistic muscle group function is prevalent [12]. Unfortunately, CS ratios exhibit a broad range across studies (flexion/extension range from 0.4 to 0.74, lateral bending/extension range from 0.6 to 1.0), which hinders researchers' ability to establish expected normative values irrespective of the measurement device.

2.1.4.6 Influence of physical activity levels:

Investigations that have tested athletes and age- and sex-matched non-active subjects using the same testing apparatus have shown, unsurprisingly, that athletes are stronger. For example, Table 1 lists the results obtained by Rezasoltani et al. [49] for six junior national level hockey players and by Valkeinen et al. [59] for age-matched, healthy subjects. In both studies, the same measurement device was used. Extension forces were found to be approximately 20% higher for hockey players. Another related study compared cervical extension strength of wrestlers and judo participants using the same testing device [58]. The results of this study showed that wrestlers have greater CS extension capabilities, which could point to adaptations and training-specificity demands of each respective sport.

2.1.4.7 Influence of symptomatic status:

A major portion of the literature concerned with quantification of CS is devoted to the evaluation of subjects exhibiting neck or head pain. All studies point out that CS values are lower in subjects suffering or having a history of neck injury, and that weakness seems to be manifested across all movement directions, irrespective of the location or nature of the injury [4,7,15,19,22,55]. Naturally, the results obtained might be severely confounded by the presence of pain or by the fact that subjects are apprehensive in exerting maximal efforts. As such, the decrease in CS scores might not be reflective of muscular strength dysfunction *per se*. It should be noted that CS conditioning has been advocated as a conservative treatment option for relieving painful symptoms and restoration of normal function [4,7], and that several investigations report improvements in these latter outcome measures [11,37,44].

2.1.4.8 CS measures other than peak force or moment:

The outcome measure used to assess CS in the vast majority of studies is the peak force or moment recorded during maximal exerted efforts. The paucity of basic information regarding other CS time-dependent measures is perplexing. Specifically, only two investigations have reported upon CS strength measures other than peak force or moment. The first study quantified the total power achieved at 0.2 seconds in elite rugby players performing isokinetic maximal efforts at 30°/sec [42]. In another study, and one that is more relevant to the aims of the current investigation, Valkienen et al. [59] quantified the maximal rate of force development in isometric extension and flexion in groups of men and women divided according to age (18-26 yrs., 30-37 yrs., and 45-55 yrs.). Valikeinen et al. [59] report that men produced higher RFD values and that in both genders higher RFD values in extension were obtained by the youngest age group. Interestingly, no differences were seen in RFD obtained in flexion across gender-matched age

groups, nor were differences observed in peak force values. There are several drawbacks to the Valkienen et al. study [59], which otherwise present novel and important findings. The first relates to the small sample size in each gender and age group (n ≤ 10). Another drawback is that the method used to calculate RFD is not specified. Lastly, only extension and flexion efforts were quantified.

2.2 Reliability of Cervical Strength Measurements

Reliability is defined as the ability to reproduce scores on repeated testing sessions performed under the same conditions [5]. The topic of reliability in measures related to human movement performance has been heavily debated, with differences of opinion regarding which statistical methods are most appropriate. However, current consensus regarding the assessment of reliability require that three separate components be taken into account: 1) the evaluation of systematic change in performance between testing sessions; 2) the magnitude of within-subject variation in scores; and, 3) test-retest correlation which indicate whether rank-order is maintained between subjects [5,14,25,53,62,63].   Whilst taking these latter components into account, recent reviews have criticized the issue of reliability assessment in CS measurements [15,55]. To demonstrate some of the concerns expressed, Table 2 presents selected reliability measures reported in the literature for CS. The first noticeable problem is that some investigations simply do not report upon the reliability of their instrument and experimental protocol. Of particular notice is the study of Vasavada et al. [60] whose testing device was used as the basis for the design of the testing apparatus used in this investigation.

Possibly the harshest criticisms relate to the reliance of several investigations on relative reliability indices (Intraclass correlation coefficients (ICCs) or Pearson's *r*) as the main or sole outcome measures. Although relative indices are useful for the assessment of reliability, several

shortcomings related to the spread of scores hinder its usage as a sole reliability measure. Specifically, if subjects exhibit large score heterogeneity, ICC or r values may indicate that a performance measure is highly reliable. For example, Peolsson et al. [43] calculate ICC scores of CS measures by taking into account the scores of both males and females. As previously documented, females are weaker than males [4,7,8,21,31,34,56], and this may have inflated the ICC scores due to a wide score range. In opposition, if subjects are extremely homogenous score-wise, then even small variations in performance between subjects may result in low ICC values [63]. This may happen, for example, if subjects exhibit the same level of physical conditioning. Another aspect that is problematic in the use of Pearson r and some ICC models is that they do not take into account the variance in scores associated with systematic error. Wier [63] suggests that a decision regarding which ICC model to use should be done after assessing whether systematic bias is present in the data. This is usually accomplished by evaluation of either the results of a paired t-test (in case of only two testing sessions), or the results of the repeated measures analysis of variance (ANOVA) in multiple testing sessions (see below).

The change in mean score values between testing session may be a result of both random and systematic changes in performance. With regards to the latter, systematic changes between testing sessions has been implicated to occur due to learning effects, insufficient recovery time, alteration of motivational factors, and mechanical error of the testing apparatus [2,5,25]. Measurements of maximal CS efforts are challenging from this perspective, as subjects are usually unaccustomed to maximal efforts. In addition, Strimpakos and Oldham [55] argue that examiners may hesitate in instructing subjects to exert maximal efforts due to fear of injury, which may lead to inconsistent scoring in multiple testing sessions. Previous investigations concerned with reliability of CS measures have shown that differences in means are apparent between the first and second testing sessions, but that differences are diminished between the

second and an added third testing session [21,55]. This has led the investigators to recommend performance of a familiarization session prior to CS testing, which is also a general recommendation for all forms of strength testing [12]. The difficulty with this recommendation is that when testing a large number of subjects, it is often not feasible to perform a familiarization session due to financial and time constraints. It might be argued that in a population regularly performing physical activities, a well-planned pre-testing session might suffice to eliminate subject apprehension, as well as shorten the learning time associated with performance of a novel task. With regards to statistical methods for evaluating differences between means, it has been recommended that in a simple reliability study incorporating only two testing sessions, the difference between means be reported as a percentage with accompanying 95% confidence intervals (CI) [2,5,25]. This reporting method allows for the assessment of the likely range of difference scores, as well provides information regarding the statistical significance of the results. It should be noted that a difference in the mean by itself cannot be used to assess the reliability of a measure. Atkinson and Neville [2] provide excellent examples showing how large intra-subject differences between test sessions may exist, yet the mean difference scores between sessions are small and insignificant. As such, and similarly to the recommendations regarding the use of retest correlations, the mean difference between scores should be evaluated in conjunction with other reliability indices.

Perhaps the most important reliability component that should be assessed is the within-subject variation in scores between testing sessions. This component is usually seen reported as the standard error of the measurement, either in the original units of measurement, or expressed as a coefficient of variation. The latter representation is more easily interpretable, as well as allowing for comparison of results of different investigations because it is dimensionless [5,14,25]. The importance of quantifying the random variation in performance is critical because

its magnitude directly affects the ability to detect meaningful changes in performance [2,5,14,25]. For a single subject, the variability in performance across multiple testing sessions is simply the standard deviation of the scores obtained. This has also been referred to as the typical error in performance [25]. Although it is appealing to calculate the mean typical error for a group of subjects in order to attain an estimate of the within-subject variability, this practice has been discouraged due to possible bias resulting from changes in the mean between trials [25]. The typical error for a group of subjects is calculated by dividing the standard deviation of the difference scores between the first and second testing sessions by the square root of two [25]. The interpretation of the typical error is similar to that of a standard deviation; it is simply the variation in scores that can be expected to be seen in testing sessions. A related within-subject variation measure is the limits of agreement (LOA) [2], which represents the 95% likely range of a subject's difference scores [2]. The computation of the LOA using the typical error is similar to what is known as the "smallest detectable difference" that must be exceeded in order for clinically meaningful changes in performance. [2,14]. The computation involves multiplying the typical error by 2.77 [25], and as such, it is easily observable that if the typical error is large, it hinders small changes in performance being declared as "real ones" at the 95% confidence level [2,25]. Thus, it would seem that the natural aim would be that within-subject variation be as small as possible.

Another criticism in regard to the reliability of CS measurements relates to the presence of heteroscedasticity (non-uniform residual distribution) and non-normality of distribution that is often not taken into account prior to calculations, although all of the reliability indices discussed above are based on these underlying assumptions [2,14,25]. In addition, Dvir [14] argues that the presence of heteroscedasticity in muscular strength measures is very common and that, if present, transformation of data (i.e. via logarithmic transformation) is required. It should be noted that

transformation of data may result in inadvertent data shifting, which would subsequently require the re-evaluation of the transformed data. This transformation step is rarely done in practice. A related problem is the fact that the majority of investigations utilize a relatively small sample size when evaluating reliability, which makes the assessment of heteroscedasticity and normality difficult.

Whilst the latter paragraphs have highlighted the criticism towards methodological approaches utilized in the assessment of CS reliability, there is also agreement that investigators that have employed adequate experimental designs and procedures have found peak force or moment values to be reproducible between testing sessions [15,55]. Reliability results of other time-dependent measures (rate of force development, force at a preset time, etc) are non-existent.

Table 2.2: Selected reliability indices reported upon for cervical strength testing

Author (s)	n	Type and Procedure*	Ext	Flex	Lat Bend	Rotation
Cagnie et al [7]	12, mixed sex	Intra, 2 tests a week apart.	SEM 2.36 Nm ICC 0.94 (CI 0.85 -0.98)	SEM 1.03 Nm ICC 0.96 (CI 0.91-0.99)		
Gera (I) [22]	25 male, age range 24-43	Intra, 2 tests 1-4 weeks apart	SEM 32.2 N ICC 0.94	SEM 15.3N ICC 0.96	SEM 31.8-33.2N ICC 0.89-0.90	
Garces et al (II) [21]	22, mixed sex	Intra, 2 tests a week apart	CV <15% Paired t NS Pearson r=0.95	CV < 15% Paired t NS Pearson r=0.95		
Strimpakos et al (III) [56]	33, mixed sex	Intra, 3 occasions 5-8 days apart	SEM 20.9 N ICC 0.94	SEM 18.9 ICC 0.97	SEM 17.0-19.3 ICC 0.93-0.97	SEM 1.4-1.5 ICC 0.97-0.9
Suryanarayana & Kumar [57]	10, mixed sex	Intra, 2 tests a week apart	Paired t NS Pearson r =0.88	Paired t NS Pearson r =0.7		

Author (s)	n	Type and Procedure*	Ext	Flex	Lat Bend	Rotation
Ylinen et al [64]	33, mixed sex	Intra, 3 different days	paired t NS SEM =10.6-11.8 N Pearson r =0.96 -0.98	Paired t NS SEM =7.1-7.7 N Pearson r =0.94 -0.98		paired t NS SEM 0.48 -0.62 Nm Pearson r = 0.89 -0.96
Rezasoltani et al [47]	35, mixed sex	Intra, 2 successive days	Paired t NS CV 6.2% SEM 3.9 N ICC 0.97	Paired t NS CV 12.4% SEM 4.9 N ICC 0.94		
Oksanen et al [41]	Total of 89 subjects, mixed sex, mixed symptomatic status	Intra, 2 tests, same day	CV range 2.3-3.7% ICC range across groups 0.98-0.99	CV range 2.7-3.5% ICC range across groups 0.98-0.99		CV range 1.6-3.5% ICC range across groups 0.98-0.99
Netto & Burnett [40] (IV)	5 male	Intra, 2 tests seperated by 2 weeks	%SEM 1.7 ICC 0.95	%SEM 5.5 ICC 0.85	%SEM 2.8 ICC 0.95	
Chiu& Sing [9](V)	25 healthy mixed, 21 neck pain	Intra, 2 test session 2-3 day apart AFTER practice session	ICC across groups 0.97-0.99	ICC across groups 0.98-0.99	ICC across groups 0.92-0.97	
Seng et al [50] (VI)	10 males	Intra, 2 tests, seperated by one week	ICC 0.93	ICC 0.82	ICC 0.83-0.91	

Results for lateral bending and rotations reported as range observed for left and right efforts.
NS = not statistically significant. **Intra** = intra examiner reliability. **SEM** = standard error of measurement. **CV** = coefficient of variation (%). **CI** = 95% confidence interval.
* Only intra-examiner reliability indices presented. Some studies also examined inter-examiner reliablity.
(I)   Original units in kgf.
(II) Results for tests 2 vs. 3. CV in extension values for 0 and 5 degrees EXT. CV for 10 degrees EXT = 26%.
(III) Only sitting position presented. Standing position proved to be more reliable. Only results for tests 2 vs. 3 presented.
(IV) Only maximal efforts, between day measurements using isokinetic device presented. Also preformed sub-maximal efforts with different testing apparatus.
(V) Chiu & Sing [9] also present data for retraction and protraction.
(VI) Also report intra-day reliability. Mean coefficient of variation across days and direction is 8%.

2.3 Electromyographic Measurements of the Neck Musculature

In light of the relatively high prevalence of musculoskeletal pain and discomfort documented for the head and neck regions, it would seem that the use of EMG for assessment of neck muscle function may provide valuable additional information. There is also agreement that

neck EMG recordings pose several methodological challenges, specifically with respect to which muscles can be measured non-invasively, as well as the need for standardization of electrode location sites [54]. Recent investigations have attempted to address these issues and will be briefly reviewed as they provide the procedural basis for the current investigation.

Falla et al. [17] report on the reliability of selected EMG indices of the sternocleidomastoid (SCM) and anterior scalene (AS) during 15s isometric efforts at 50% of peak force capability in flexion on 3 separate testing days. The authors state that the values of the relative and absolute reliability measures (ICC and normalized standard error of measurement) of the slope and initial average rectified values for the SCM and AS justify the use of EMG in clinical assessment using the proposed protocol. The acceptable level of reliability demonstrated in this investigation may be attributed in large part to the standardization of electrode placement relative to the muscles' innervation zone. These specific locations were quantified for the first time using a procedure involving recordings from a linear array of eight electrodes and observation of phase reversal and diminished signal amplitude [17,18]. The investigations conducted by Falla et al. [17,18] are noteworthy, especially since detailed guidelines regarding electrode placement sites are provided. Even so, it should be noted that severe criticism has been voiced by Kennedy and Mercer [30] regarding the availability of the AS for surface EMG recordings using the techniques prescribed by Falla et al. [17,18]. Specifically, these authors demonstrate, using cadaver material and anatomical imaging, that the location of the AS is deep to the clavicular head of the SCM, and that manual palpation of the AS using the techniques described by Falla et al. [17,18] cannot be done.

With regards to the dorsal neck muscles, Joines et al. [27,28] investigated, by means of ultrasound, whether the splenius capitis (SpL) and semi-spinalis capitis (SEMI) muscles are accessible for surface EMG recordings in self-selected neutral head postures. The results of their

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investigation showed that SEMI is largely inaccessible; refuting statements made earlier by Keshner et al. [31] regarding the availability of this muscle for surface EMG recordings. In addition, even in those subjects where SEMI could be measured, it would seem that preparation of the electrode sites would require a significant portion of the subjects' hair to be shaved. This would entail a lengthening of the pre-testing phase and, as well, would require extremely generous subject cooperation. On the other hand, SpL was found to have a larger and easily accessible gap between the SCM and upper trapezius, confirming results of an earlier study by Quessier et al. [46]. It should also be noted that Joines et al. [27,28] argue that, with careful placement, the electrode pair for the SpL may be positioned with no overlapping of the SCM , and that the electrode site may be referred to as a single muscle, rather than given a global description such as "neck extensors" as proposed by Mayou-Benhamou [38].

Based on the results of these recent studies, it was decided that the bilateral recordings of the SCM, SpL and upper trapezius (TRP) will be recorded in this investigation. With regards to the TRP, although its anatomical attachments at the base of the skull suggests that it acts as a primary head extensor, it has been argued that it acts only as a synergist to the action of deeper agonistic muscles [31]. Even so, recording the EMG from the TRP are made in this investigation as we hypothesize that this muscle may play a significant role in concussion prevention. This role for the TRP might be possible merely because of its relative large size in comparison to the other muscles of the neck. In addition, the TRP is relatively easy to condition using standard resistant exercises.

2.4 Reliability of Surface Electromyography of the Neck Muscles

With regards to the reliability of amplitude-based EMG measures, Oksanen et al. [41] investigated the intra-examiner reliability of the average full-wave rectified EMG signals of the

SCM and cervical erector spinae muscle group in healthy adolescents and those suffering from headaches. Oksanen et al. [41] report ICC values of 0.95 to 0.99 with low coefficients of variation ranging from 4.9% to 10.1% for participants performing maximal voluntary contractions in extension, flexion and bilateral rotation. The excellent reliability reported in this study may in large part due to the fact that testing sessions were separated by only one hour, and it is suspected that the electrodes were not repositioned between testing sessions. In addition, ICC calculation incorporated both healthy and symptomatic subjects, and thus may have inflated the ICC values. It should also be noted that EMG values were not normalized in this investigation, which might be questionable if measurements were performed on different days.

Falla et al. [17] report the repeatability of the signal mean frequency, average rectified value, and conduction velocity of the SCM and AS in nine healthy volunteers performing sub-maximal flexion efforts on three non-consecutive testing days. The authors stated that acceptable reliability was found for the initial value and slope of the average rectified value of the SCM (ICC > 0.65) and the initial mean frequency and slope of the average rectified value of the AS (ICC > 0.70 and ICC > 0.75, respectively). The normalized standard error of the mean for these variables ranged between 2.8% to 7.8%. As previously discussed, the validity of the recordings obtained for the AS has been questioned [30].

A specific variable to be measured in this investigation is the activity onset of selected neck muscles relative to the onset of force rise, referred to as the electro-mechanical delay (EMD). Although EMD has been the subject of numerous investigations, there is a surprising lack of information regarding its reliability. Most recently, Howatson et al. [26] conducted a study where the EMD of the dominant and non-dominant biceps brachii was recorded over five consecutive days in isometric and isokinetic contractions. These investigators report a measurement precision using the coefficient of variation (CV), which ranged between 3.1% and

6.5%. The ICC scores for isometric contractions were 0.55 and 0.79 for the dominant and non-dominant arm, respectively. In another recent study, Minshull et al. [39] report a %SEM of 15.9 with an ICC value of 0.64 for EMD of the biceps femoris when measured isometrically in 12 adults on three testing occasions separated by at least three days. In an earlier study, Zhou et al. [67] briefly mention that ICC values of 0.97 to 0.98 were obtained for EMD measurements of selected knee extensors in six subjects performing isometric contractions in a test-retest scheme separated by one week. However, no estimates of measurement precision are given.

Interest in the EMD has also been seen in studies concerned with the prevention of ankle sprains: Eechaute et al. [16] report SEM values of 2.7ms for the EMD of the peroneus longus during sudden ankle inversion tests that were separated by one week, with an ICC value of 0.17, whilst Grosset et al. [24] report that ICC of 0.96 and CV of 0.26% for EMD of the gastrocnemius in quick release tests.

It should be noted that no investigations were found that assessed the reliability of the muscle activation onsets in neck muscles.

2.5 Summary

Evaluation of current knowledge regarding CS and neck EMG measurements suggests that many aspects related to the function of the neck musculature have not been explored. This stems primarily from limitations of current testing apparatus used to measure CS, as well as the inherent limitations of EMG. In addition, and with regards to the aim of the current investigations, many outcome variables that may be deemed important for athletic populations have not been reported in the scientific literature: as well, the majority of investigations report only movement efforts along one or two directions. In addition, reliability of CS and neck EMG remains a topic that has not been adequately addressed.

The following two chapters address many of the limitations mentioned above: Chapter 3 elaborates upon the reliability of isometric neck strength measures that are relevant to concussion prevention in athletes, whilst chapter 4 addresses the between-day reliability of activity onset of selected neck muscles during performance of maximal isometric efforts. In a concluding chapter, the importance of these findings will be integrated with the current scientific literature regarding CS measurement techniques.

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Chapter 3

Reliability of Isometric Neck Strength Measures: Relevance to Concussion Prevention in Athletes

3.1 Abstract

Background: Strengthening of the neck muscles may play a role in reducing the risk of sport related concussions. However, the ability to reliably measure force-time based variables that might be relevant for this purpose has not been addressed. The purpose of this study was to assess the between-day reliability of discrete force-time dependent variables of neck muscles during maximal voluntary exertions in isometric extension, flexion, protraction, and lateral-bending.

Methods: Twenty six highly physically active males were tested on two occasions, 7-8 days apart, using a self-fabricated testing apparatus. Dependent measures were peak force (PF), rate of force development (RFD) and time to 50% of peak force ($T_{50}PF$). Reliability indices calculated for each variable were the difference in scores between the two testing sessions, with corresponding 95% confidence intervals, the typical error of measurement (TE), and intraclass correlation coefficients (ICCs)

Results: No evidence of systematic bias was detected for the dependent measures across any movement direction: percent difference range of -1.8 to 2.7%, with 95% confidence interval range below 10% and overlapping zero. TE was lowest for PF, with a range of 2.4% to 6.5% across all testing directions, followed by RFD (4.8% to 9.0%) and $T_{50}PF$ (7.4% to 9.3%). ICCs ranged from 0.90 to 0.99.

Conclusion: Discrete variables representative of the force generating capacity of neck muscles under isometric conditions can be measured with an acceptable degree of reliability. This has possible applications to investigating the role of training programs in reducing the risk of concussions in collision sports.

3.2 Introduction

Concussions sustained in collision sports are a serious health problem that carries the potential for long-term impairment in brain function [1-3]. Researchers have proposed that improving the strength capacity of neck muscles could reduce the risk of impact related concussions, and therefore should be incorporated as a component of a well-balanced athlete training program [4-6]. The work by Viano et al. [6] on the mechanics of head responses to impacts in football theoretically supports this putative role of neck muscle strengthening: using a modeling approach, these investigators demonstrated that progressive increases in neck stiffness prior to impact reduced both the change in velocity and peak acceleration of the head during simulated head-to-head collisions in football. Therefore, neck stiffness could directly influence the mechanical response of the head to impact by limiting the transmission of injurious forces to the brain.

The results presented by Viano et al. [6] are attractive from an interventional perspective as the inclusion of neck muscle conditioning in regular training regimes is simple and cost effective. However, two fundamental issues related to the measurement of neck muscle capacity need to be considered as a pre-requisite to prospective and interventional studies. The first issue concerns the selection of appropriate outcome variables to assess neck muscle function in the context of concussion prevention. To date, the vast majority of investigations on isometric neck testing have only reported upon peak force or moments as outcome [7-14]. However, peak force may not be completely relevant to investigating the role of neck muscle strength in concussion prevention as in real situations of play, an athlete may not attain maximal muscle force prior to contact. This may be due to poor relative awareness of the impending collision and a limited time to generate sufficient muscle force prior to impact. In this respect, quantifying the early force generation capacity of neck muscles would be more meaningful, as these variables would provide insight

into the short-term damping response of the neck. Such relevant variables would include, in addition to peak force, the rate of force development and the time needed to reach a percentage of peak force that would result in meaningful increases in neck stiffening.

The second issue concerns our ability to reliably measure these aforementioned variables. Dvir and Prushansky [15] have recently raised the issue of reliability in measurement of neck muscle function to be a concern in interpretation of results from different studies. Specific concerns include the use of heterogeneous study populations and over reliance on interpretation of relative reliability indices (i.e. Pearson's *r* or intraclass correlation coefficients).

The purpose of this study was to determine the retest reliability of peak force, rate of force development and time to reach 50% of peak force in athletes performing maximal isometric neck muscle exertions in five different directions. The results will be relevant to investigating the potential role of neck muscles in modifying the mechanics of the head to imposed loads, as well as to quantifying the effects of strengthening programs on the force generating capacity of the neck.

3.3 Methods

3.3.1 Subjects

Male athletes were recruited within the University community. Prospective subjects were screened using a self-report questionnaire on the set of exclusion criteria proposed by Sommerich et al. [16] (Appendix B): neck injury or pain; head injury; recurrent episodes of fainting or dizziness; surgical interventions to the head; neck or shoulder regions; high blood pressure, current use of medications; and high risk for carotid and coronary artery disease. Twenty-six subjects were tested (mean (SD)): age 21.6 (2.1) years; weight 81.6 (9.9) kg; height 1.85 (0.09) m; head circumference 0.58 (0.01) m; and neck circumference 0.39 (0.02) m. All subjects

participated in regular physical activity 4 to 8 times per week in team or individual sports at the competitive university, national, or elite level; none of these sports involved specific conditioning of the neck muscles as part of their training routines. The methods and procedures were approved by the University Research Ethics Board, and subjects provided written informed consent.

3.3.2 Instrumentation

A custom-built neck strength testing device was developed based on the work of Vasavada et al. [14] The device, shown in Figure 1 and Appendix H, includes the following key features: Measures of exerted neck forces are recorded using a six degree of freedom load cell (model MC5-6-2500, AMTI, Watertown, MA, USA), mounted on a three-point base of support metal frame that is secured to a heavy plywood platform which is immovable during testing due to the combined mass of the device and the subject. The metal support frame is further stabilized by a horizontal metal brace that is bolted to the wall; this brace eliminates any bending of the frame as subjects exert maximal forces. Coupling of the subjects' head to the load cell is achieved by attaching an ice hockey helmet and face cage (model 8500, Bauer Nike, St. Jerome, Quebec, Canada) to a semi-spherical aluminum support structure that is secured to the load cell using eight counter-sunk threaded bolts. The Bauer helmet was selected for its adjustability features that reduce the amount of relative motion between the head and the helmet during testing. The helmet shell is affixed to the aluminum support via a three-point attachment system. The two lateral attachments are fixed in position, while the third vertical attachment allows placement of the head in different positions in the sagittal plane within a 70 degree range of movement. During testing, subjects sit in a steel-frame chair that is attached to the supporting frame by a metal brace. The chair is fitted with a four-point seatbelt restraint system that is used to stabilize the subjects' shoulders and lower torso. An additional padded chest strap is used to stabilize subjects' upper

torso. The height of the load cell and the distance of the chair relative to the load cell are fully adjustable to accommodate subjects' of different dimensions.

The accuracy of the load cell factory-calibration specifications was verified prior to testing using known weights between 0.9 and 8.9 kg placed along the orthogonal directions of the load cell. The root mean square error for all force and moment channels was less than 1 N and 0.1 Nm, respectively, and the coefficient of determination values (R^2) were > 0.99, indicating a linear response within the measurement range (see Appendix C).

Figure 3.1: Custom-built testing apparatus.

3.3.3 Procedures

The same investigator performed all measurements and provided all verbal instructions. At the beginning of each testing session, subjects completed warm-up exercises consisting of: (1) active range of motion of the neck with passive stretching at end-range, (2) performance of 3-5

self-resisted sub-maximal isometric exertions of the neck muscles in the five directions of testing (extension, flexion, protraction, and left and right lateral bending), and (3) 1-2 self-resisted maximal exertions. Subjects were asked if any of these maneuvers caused pain prior to proceeding with the experimental protocol.

Subjects were fitted with the hockey helmet, seated in the device, and stabilized with the restraint system. Subjects were asked to assume a comfortable, neutral position of the head and neck, and this self-determined neutral position was recorded using a 3-Space Isotrak digitizer system (Polhemus Navigation Sciences, Colchester, VT, USA). The helmet was firmly coupled to the fixed frame, the neutral posture was again digitized, and the position of the head within the fixed frame was adjusted, as needed, to be within 2 degrees of the previous recorded neutral position. Subjects were instructed on the standard positioning during testing: both hands on the thighs, and feet resting on a cardboard box. The latter enabled the examiner to easily detect both audibly and visually whether the legs were contributing to the recorded force, as pushing down on the box would collapse it and pushing to the sides would translate the box across the surface. When this occurred during testing, the trial was discontinued and repeated after a 30-second rest period.

Thereafter, subjects performed 2-3 sub-maximal force exertion trials in each of the 5 directions of muscle testing, followed by 1 maximal voluntary muscle exertion (MVE) trial in each testing direction. Subjects were again questioned to ensure that none of the maneuvers produced neck pain. During this familiarization session, a standardized set of instructions was used to guide subjects on correct performance of the task, as well as on the use of the real-time visual feedback provided: subjects were instructed to reach their maximum force as fast as possible and then to hold this force level until the end of the trial. In order to facilitate understanding of these

instructions, the examiner showed the participants force-time curves of their practice trials, and compared them to the shape of the expected isometric force-time curve (e.g. figure 2). Specifically, the examiner commented on whether the effort seemed "graded"

Subjects performed 4 MVEs for each of the 5 test directions: extension, flexion, protraction, and left and right lateral bending. Each trial lasted 4 seconds in duration, with a 30 second rest period between trials. The order of directions for MVEs was fully randomized both within and between subjects. Visual feedback on performance and consistent verbal encouragement were provided throughout the testing session.

3.3.4 Retest Procedures

To determine between-day retest reliability, subjects completed a second testing session, 7-8 days later with care being taken to schedule it at the same time of day to control for diurnal effects. The subjects were positioned in the device according to the set-up on the first testing day (see Appendix F). The order of movement directions was the same as performed in the first testing session.

3.4 Signal Acquisition and Processing

The force signals were amplified (Modular 600 multi-channel amplifier, RDP Group™, Pottstown, PA, USA; ±10 V peak-to-peak range, frequency response 0-1 kHz, CMRR 110 dB at 60 Hz, input impedance 100MΩ) and sampled using a 16-bit Analog to Digital converter (NI PCI-6036E, range of ±5V) at 2048 Hz. The signal acquisition was controlled using custom-software written in Labview™ version 8.6 (National Instruments Inc., Austin, TX, USA).

Signals were zero offset and low pass filtered using a 2nd order, dual pass Butterworth digital filter with 15Hz cut-off frequency. The onset of force development was defined as the instant the force-time curve exceeded a value of 2 standard deviations above baseline levels, and

remained above this value for a period of 100 msec. All onset instances were verified by visual inspection. Thereafter, the following dependent variables were extracted from each individual force-time curve (Figure 2): (1) Peak force (PF, units N), defined as the maximum force value produced over the trial;(2) rate of force development (RFD, units Ns^{-1}), defined as the maximal value of the slope of the force time curve; and calculated using a continuous 50 ms sliding window from onset to peak force, and (3) Time to 50% of peak force ($T_{50}PF$, units ms).

Figure 3.2: Isometric force-time curve identifying the three extracted measures: peak force (PF), rate of force development (RFD) and time to 50% peak force (T50PF).
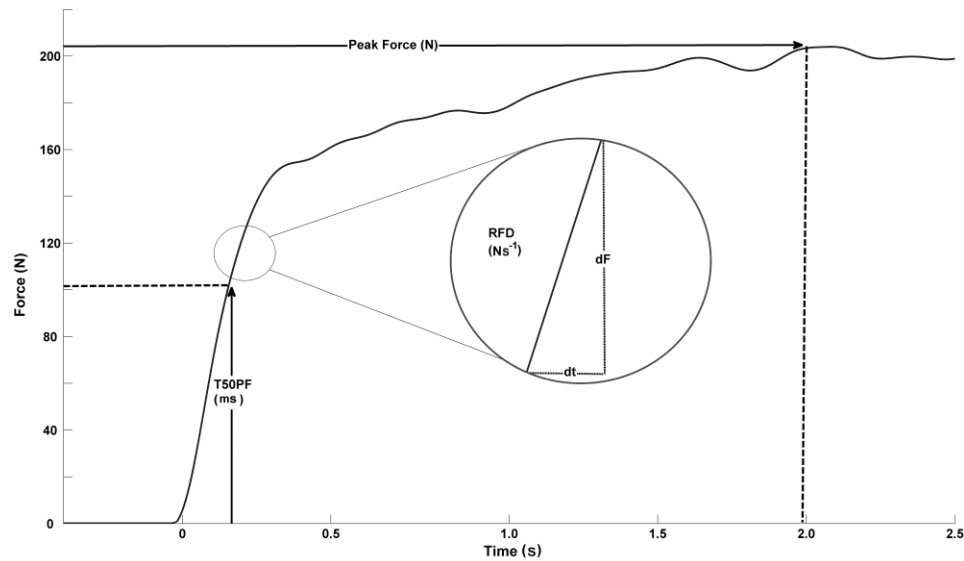
3.5 Statistical Analysis

For each outcome variable, the 3 best scores (e.g. highest PF and RFD; lowest $T_{50}PF$) in each of the 5 testing directions were used to calculate an average individual score. These average scores were subsequently used in the following analyses of reliability.

3.5.1 Data Inspection

Prior to calculation of reliability components, all data were evaluated for non uniformity in error distribution (heteroscedasticity) and normality of distribution (see Appendix D). Heteroscedasticity was assessed using the method proposed by Atkinson and Nevill [17], in which the absolute difference scores between day 1 (D1) and day 2 (D2) for all dependent variables were plotted against their respective D1 and D2 mean values, and the Pearson product moment correlation coefficient calculated for each of these relationships. Normality of distribution was assessed for each dependent variable by visual inspection of histogram plots of the difference scores between the two testing sessions and by Shapiro-Wilks normality tests ($p \leq$ 05). Based on the results, all data were log transformed and subsequently multiplied by 100 [20,21]. Log transformation may produce inadvertent shifts in the distribution of the data and as such, the normality of the distribution of the log-transformed data was evaluated again using the methods described above.

3.5.2 Reliability Measures

For indications of systematic bias, the difference in average scores between testing sessions (i.e. Day 2 score - Day 1 score) were computed along with the corresponding 95% confidence intervals [18,19,21]:

$$\text{Percent Difference (\%)} = 100 \ e^{x/100} - 100, \tag{1}$$

where e is the base of the natural logarithm and x is the mean of the group log transformed difference values.

Measurement precision was assessed using the typical error (TE) of measurement, expressed as a coefficient of variation (CV_{TE}) to permit comparison across the different measures [18,19,21]:

$$CV_{TE}(\%) = 100 \ e^{TE/100} - 100 \tag{2}$$

We also calculated the smallest detectable difference value (SDD), which is used to determine the smallest change necessary for declaration of statistically significant differences between measurements from the two testing sessions [18]. The SDD was calculated by multiplying TE by 2.77 [19].

Retest correlation was calculated using intraclass correlation (ICC) coefficients (model 3,k) [22].

All statistical analyses were performed with SPSS (version 15; SPSS Inc., Chicago, Illinois, USA), Excel 2007 (Microsoft Corp., Redmond, WA, USA) and Matlab (version 7.5; MathWorks Inc., Natick, MA, USA).

3.6 Results

Group means (1 SD) for PF, RFD and T_{50}PF for D1 and D2, and the between-day percent differences (95% CI) are reported in Table 4. For all dependent variables, the between-day percent difference across all testing directions ranged from -1.8% to 2.7%, with 95% CI ranges below 10% and overlapping zero, indicative of no statistically significant differences in measurement between the two testing sessions.

The typical error (CV_{TE}), retest correlations (ICC) and smallest detectable difference (SDD) values are reported in Table 5. CV_{TE} values for all dependent variables were less than 10%, with values for PF (2.4-6.3%) across all movement directions being slightly lower than values obtained for RFD (4.8-9.0%) and T_{50}PF (7.4-9.3%).

Retest correlations were high for all variables, with ICC values across all movement directions ranging between 0.90 and 0.99. Finally, SDD values were lowest for PF (6.6-17.4%), followed by values for RFD (13.3-25.0%) and T_{50}PF (19.8-25.7%).

Table 3: Group mean (1 SD) of force measurements at D1 and D2 and corresponding between day differences in measurements (95% CI)

Variable	Movement	D1	D2	% Difference
Peak Force (N)	Extension	253.2 (48.0)	250.7 (44.2)	-0.8 (-3.2 to 1.6)
	Flexion	149.7 (27.0)	152.0 (30.2)	1.5 (-1.9 to 5.1)
	Protraction	155.0 (33.3)	157.7 (35.4)	1.5 (-1.9 to 5.0)
	Left LB	157.2 (37.6)	158.1 (37.2)	0.6 (-0.7 to 2.0)
	Right LB	161.8 (38.3)	158.9 (33.1)	-1.3 (-4.6 to 2.1)
RFD (Ns^{-1})	Extension	1614 (491)	1605 (449)	0.2 (-3.7 to 4.4)
	Flexion	865 (290)	860 (312)	-1.1 (-5.9 to 3.9)
	Protraction	1012 (350)	996 (311)	-0.6 (-4.2 to 3.2)
	Left LB	1231 (462)	1261 (458)	2.7 (0.0 to 5.5)
	Right LB	1351 (466)	1335 (443)	-0.8 (-4.0 to 2.5)
$T_{50}PF$ (ms)	Extension	135 (29)	132 (23)	-1.8 (-6.3 to 3.0)
	Flexion	146 (65)	148 (61)	1.5 (-3.6 to 6.8)
	Protraction	142 (40)	138 (32)	-1.3 (-5.7 to 3.4)
	Left LB	138 (28)	137 (27)	0.2 (-3.7 to 4.2)
	Right LB	148 (40)	146 (42)	-1.1 (-5.7 to 3.8)

D1 = Results obtained in day 1. **D2** = Results obtained in day 2. **LB** = lateral bending. **RFD** = Rate of force development. $T_{50}PF$ = Time to 50% of peak force

Table 4: Reliability indices for force-time curve based variables

Variable	Effort	CV_{TE} (95% CI)	$ICC_{3,3}$ (95% CI)	SDD (%)
Peak Force (N)	Extension	4.4 (3.5 to 6.2)	0.97 (0.94 to 0.98)	12.2
	Flexion	6.3 (5.0 to 9.1)	0.94 (0.88 to 0.97)	17.4
	Protraction	6.1 (4.9 to 8.8)	0.96 (0.91 to 0.98)	17.0
	Left LB	2.4 (1.9 to 3.3)	0.996 (0.992 to 0.998)	6.6
	Right LB	6.1 (4.9 to 8.8)	0.97 (0.93 to 0.98)	17.0
RFD (Ns^{-1})	Extension	7.3 (5.9 to 10.6)	0.97 (0.94 to 0.99)	20.3
	Flexion	9.0 (7.3 to 13.3)	0.97 (0.94 to 0.98)	25.0
	Protraction	6.7 (5.4 to 9.7)	0.98 (0.95 to 0.99)	18.6
	Left LB	4.8 (3.9 to 6.9)	0.99 (0.98 to 0.99)	13.3
	Right LB	5.9 (4.7 to 8.5)	0.98 (0.97 to 0.99)	16.4
$T_{50}PF$ (ms)	Extension	8.6 (7.0 to 12.7)	0.90 (0.78 to 0.95)	23.9
	Flexion	9.3 (7.5 to 13.6)	0.96 (0.92 to 0.98)	25.7
	Protraction	8.5 (6.9 to 12.4)	0.94 (0.88 to 0.97)	23.4
	Left LB	7.1 (5.8 to 12.4)	0.93 (0.84 to 0.97)	19.8
	Right LB	8.8 (7.1 to 12.9)	0.95 (0.89 to 0.97)	24.2

LB = lateral bending. **RFD** = Rate of force development. $T_{50}PF$ = Time to 50% of peak force. CV_{TE} = Typical error expressed as a coefficient of variation. **ICC** = Intraclass correlation coefficient. **SDD** = Smallest detectable difference.

3.7 Discussion

As suggested by Viano et al. [6], strength training of neck muscles should be included in a comprehensive program to manage the risk of concussions in collision sports. This recommendation is based on predictions from a computational model and requires verification through controlled, population-based studies. A first step towards this goal is to obtain measurements of neck muscle forces that are both reproducible and meaningful to the training of athletes involved in collision sports. Our study introduces a self-fabricated strength testing apparatus that is specific to sport populations and provides evidence of retest measurement reliability.

The quantification of the percent difference of measurements between testing sessions is meant to identify systematic bias that may be introduced by factors such as test-retest learning, motivational differences or insufficient recovery time [23]. In this investigation, there are no indications of systematic bias for any of the dependent variables between the two testing sessions.

The typical error is a measure of the within-subject variation in performance between testing sessions and as such provides an estimate of the precision of the measured variables. The magnitude of the typical error, therefore, directly influences the ability to detect statistically meaningful changes in performance. With respect to variables measured in this study, the low magnitude of the typical error for peak force would allow changes in performance of 20% or lower to be declared as real differences. This level of precision is well within the expected range of change in neck peak force following intervention programs, which have been reported to be between 24% and 64% [7-9].

The typical error values for RFD and $T_{50}PF$ were higher thus limiting small differences in performance, typically below 20%, to be declared as statistically significant. Improving the precision of measurement for these variables would be important to evaluate the predictions of

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Viano et al. [6], that small increases in neck stiffness have a substantive effect on damping the initial mechanical response of the head impact loads during collision. To address this issue, future research should evaluate the effectiveness of including a familiarization session prior to testing [24,25]. Even so, it should be noted that following training, RFD values obtained from different muscle groups have been documented to improve between 17% and 33% [26-28]. and that improvements are expected to be accentuated in those that are unaccustomed or untrained in this type of effort [29]. Thus, while lower typical error values for RFD and $T_{50}PF$ would clearly be beneficial, the typical error magnitudes in this study would suffice to detect changes in performance following an intervention program, especially if participants are not accustomed in performing maximal isometric exertions of the neck musculature.

Relative reliability, commonly assessed by some form of retest correlation (Pearson *r* or ICC), allows for the assessment of rank order maintenance among participants between testing sessions [19,30,31]. ICC values obtained in this study for all dependent measures are considered high and are in agreement with previous investigations that have reported this measure for peak force values of isometric neck exertions measured in sitting using fixed-frame dynamometry [13,25,32,33]. There has been some debate as to the usefulness of the ICC in the assessment of reliability, particularly due to the inherent sensitivity of the ICC to score heterogeneity [15,19,23]. In our study, we did try to minimize the effect of this confounding factor by utilizing a sample that was highly homogenous in terms of physical characteristics and amount of athletic training. In addition, ICC values may be affected by the inclusion of systematic error components, as well as by the use of single trial values versus the average of several trials [30,31]. In our analysis, we used the 3,k ICC model, which does not take systematic bias into account, as there was no evidence that such error existed [31]. In addition, the use of average scores as opposed to the single best score in the ICC calculation is justified by our interest to

measure variables that would be relevant to collision sports, where athletes would be required to exert maximal muscle efforts over a short period of time but repetitively over the course of a game.

The custom-built testing apparatus used in this study addresses some of the limitations in technology and methodology regarding measurement of neck muscle capabilities [15,24]. Specifically, the device uses a commercially available hockey helmet to couple the head and neck to the load cell. This improves subjects' comfort and willingness to exert maximal excursions, especially if the helmet is routinely donned during sport participation. In addition, the use of an adjustable sport protective helmet that includes a chin strap system allows for the independent measurement of forces in the direction of flexion and protraction.

As there is no gold standard for isometric neck forces, the values obtained across the five movement directions were compared to previously reported data in healthy populations to establish baseline validity for our device. In general, our results agree with the observations made by most investigations where the largest maximal force is recorded for the extension effort [10,11,24,34,35]. In addition, lateral bending peak forces in our investigation are within 10% of those measured for flexion, which is in agreement with results of others [32,34,35]. Only one study was found that reported on neck muscular strength indices other than peak force. Specifically, Valkeinen et al. [36] report RFD values for men and women of different ages performing isometric extension and flexion efforts in a neutral head posture. When comparing our results to those of the age and sex matched group, the RFD values obtained in our study are 10% higher for flexion and 30% higher for extension. These differences might be attributed to the methods for calculating RFD values, which are not specified by Valkeinen et al.,[36] as well as to differences in physical conditioning between the two subject populations.

3.8 Conclusion

The results of this study indicate that time-based measures of the force-generating capacity of the neck muscles can be measured with an acceptable level of reliability using the standardized measurement device and testing protocol detailed in this study. The study further demonstrates that significant strength can be developed by the neck muscles in a relatively short period of time, a finding that may be relevant to the prevention of concussions. Subsequent research will focus on adapting our methods to reliably measure rotational forces of the neck, as well as to determine the responsiveness of measurements in athletes recovering from concussions and in those taking part in strength training programs as part of a multifaceted approach to the prevention of concussions.

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Chapter 4

Between day reliability of neck muscle activation onsets during performance of maximal isometric efforts.

4.1 Abstract

Background: The purpose of this study was to assess the between-day reliability of the activation onset of selected neck muscles during the performance of maximal isometric exertions in five different directions.

Methods: Twenty-one physically active males participated in two testing sessions separated by 7 or 8 days. Utilizing a custom-made testing apparatus, cervical force and surface electromyography (EMG) were recorded from the splenius capitis, upper trapezius and sternocleidomastoid muscles bilaterally during the performance of efforts in extension, flexion, left and right lateral bending, and protraction. The muscle activation and force onsets were extracted using the Teager-Kaiser Energy Operator. Reliability indices calculated for each muscle in each testing direction were: the difference in scores between the two testing sessions and corresponding 95% confidence intervals, the standard error of measurement (SEM) and intra-class correlation coefficients (ICC).

Results: Muscle activation onset values showed no evidence of systematic difference between the two testing sessions across all muscles and testing directions. The SEM for extension, flexion and lateral bending efforts ranged between 2.5 ms and 4.8 ms, indicating a good level of measurement precision. For protraction, SEM values were higher and considered to be imprecise for research and clinical purposes. ICC values for all muscles across all testing directions ranged from 0.23 to 0.79.

Conclusion: Activation onsets of selected neck muscles can be measured with sufficient

precision for the assessment of neck muscle function in athletic populations in the majority of

directions tested.

4.2 Introduction

Concussions related to sport participation are a serious health problem, for which a multi-tiered approach has been recommended. This approach focuses on reducing both the risk for concussion as well as the severity of injury during a concussive event through improvement of protective equipment [1], modification of playing rules [2], emphasis on fair play [3], and development of comprehensive guidelines regarding return-to-play decision making [4]. Neck strengthening exercises have also been recommended as part of a comprehensive sport participation program to manage the risk for concussion [5-7]. This recommendation, however, is supported only by indirect evidence from studies that have used model simulations to show that, theoretically, increases in neck stiffness would achieve substantial reductions in the magnitude of head accelerations during collisions [7-9].

A first step toward translating the predictions of these simulations to the design of evidence-based strength training programs for concussion prevention is the ability to reliably measure relevant variables of neck muscle function. In that respect, the relative activation onset of the neck muscles relative to the onset of force production may provide information regarding coordination patterns in both healthy and pathological patients. However, there is scarcity in information pertaining to activation onset of the neck musculature during performance of maximal exertions, and no investigations have addressed the reproducibility of this measure in a between-day test-retest scheme.

The purpose of this study was to assess the reliability of neck muscle activation onsets in subjects performing maximal voluntary excursions (MVE) in five different directions.

4.3 Methods

The procedures used in this investigation have been previously described in chapter 3 of this thesis. These will be briefly described briefly, along with other information pertinent to the aims of this investigation.

4.3.1 Subjects

Twenty-one athletic males participated in this study (age: 21 (1.2) years, height 1.88 (0.07) m, weight 82.6 (5.4) kg, neck circumference 0.56 (0.02) m, and head circumference 0.38 (0.02) m). Subjects were involved in physical training 4 to 8 times per week at the university, national or elite competitive levels. None of these activities incorporated specific training of the neck musculature. Prior to testing, all subjects provided written informed consent and completed a self-report medical questionnaire to screen for specific exclusion criteria suggested by Sommerich et al. [10] when testing MVEs of the neck. All methods and procedures for this study were approved by the University Research Ethics Board.

4.3.2 Instrumentation

A custom built, fixed-frame static dynamometer was used to simultaneously record force and EMG measures during MVEs of the neck muscles (Figure 1). The device consists of a load cell coupled to a semi-spherical aluminum structure used for attachment of a hockey helmet incorporating a face mask (Bauer Nike, St. Jerome, Quebec, CA). The position of the helmet in the sagittal plane as well as the height of the load cell and the distance of the chair relative to the load cell can be adjusted to accommodate individual subject anthropometrics. Subjects' perform testing while seated and firmly restrained. Surface EMG was recorded from the neck muscles using bipolar self-adhesive, pre-gelled Ag-Ag/Cl electrodes with an inner diameter of 10mm (Bortec Biomedical Ltd., Calgary, Alberta, CA). The EMG electrodes were interfaced with a Bortec AMT-8 amplifier (frequency response 10 Hz to 1 kHz, CMRR < 115 dB at 60 Hz, input

impedance > 1 GΩ, gain range 2000-5000) while subject-exerted forces were recorded using a six-degree-of-freedom load cell (MC5-2500, AMTI, Watertown, MA,USA). The load cell was interfaced with a multi-channel amplifier (Modular 600, ±10 V peak-to-peak range, frequency response 0-1 kHz, CMRR 110 dB at 60 Hz, input impedance 100MΩ) (RDP Group™, Pottstown, PA, USA). Both load cell and EMG data were sampled using a common 16-bit Analog to Digital (A/D) converter (NI PCI-6036E, range of ±5V) at 2048Hz using custom-software written in Labview™ version 8.6 (National Instruments Inc., Austin, TX, USA). The accuracy of the load cell factory-calibration specifications was verified prior to testing.

4.4 Procedures

Subjects' completed two testing sessions 7 or 8 days apart. All tests were performed at the same time of day to control for diurnal effects. Prior to each testing session, subjects completed a series of warm-up exercises involving movement of the head and neck through full range of motion, passive stretching at end range, and self-resisted sub-maximal and maximal isometric excursions in the five directions of testing (extension, flexion, left and right lateral bending, and protraction).

Surface EMG activity was recorded bilaterally from the sternocleidomastoid (SCM), splenius capitis (SpL) and upper trapezius (TRP) muscles (Figure 3). Prior to electrode placement, the subjects' skin was shaved, abraded with fine sand paper and cleansed with 70% isopropyl alcohol. For the SCM, the electrodes were placed along the sternal portion of the muscle, with the electrode centre 1/3 of the distance between the mastoid process and the sternal notch [11,12]. For SpL, the electrode centre was located at the intersection of the C7-ear line and the line of action of the SpL muscle that had been palpated during examiner-induced resistance to isometric exertions of the head in rotation [13]. For the TRP, the medial electrode was placed 2 cm lateral to the midpoint of the C4-C5 inter-spinous distance and oriented along the palpated

anterior border of the trapezius, in line with the direction of the muscle fibers [14]. For SCM and TRP, the inter-electrode distance was 20 mm, while a 12 mm distance was used for the SpL in order to minimize the chance of overlapping adjacent muscles. A common reference electrode was placed on the right acromium process (pre-gelled, Ag-Ag/Cl, 10mm inner diameter, Meditrace Model 135, Kendall, MA, USA). The electrodes were further secured to the skin using skin tape. Prior to recording, the electrodes were allowed to stabilize for 10 to 15 minutes, and tested with an ohmmeter to insure an electrode–skin impedance level of less than 10 kΩ.

Subjects were then fitted with the hockey helmet, seated and restrained in the device. The helmet was attached to the fixed frame and its position was adjusted to correspond to measurements of a self-determined, neutral posture of the head and neck recorded beforehand using a 3-Space Isotrak digitizer system (Polhemus Navigation Sciences, Colchester, VT, USA). Subjects were instructed to keep both hands on their thighs, and to place their feet on a cardboard box. The placement of the feet on the box enabled audible and visual identification of leg muscular contribution to force production during trial performance.

Subjects performed several sub-maximal practice trials in each direction in order to familiarize with the experimental task; once comfortable, subjects performed one or two MVE practice trials in each direction. For testing, subjects were instructed to develop force as fast as possible, and then to hold this force level until the end of the trial. Subjects performed four MVEs in each direction for a total of 20 trials. Each trial lasted four seconds in duration, with a 30 s rest period given between trials. When contribution by the lower extremities was detected, the trial was discontinued and repeated after a 30 s rest period. The order of directions during testing was randomized both within and between subjects. The same order of movement directions was used for the two testing sessions. Visual feedback on performance and consistent verbal

encouragement were provided throughout each testing session. All procedures were administered by a single examiner.

Figure 4.1: Electrode placement sites for the right side.

4.5 Signal Processing

Force and EMG signals were zero offset prior to processing. The first step in determination of onsets involved the computation of the Teager-Kaiser Energy Operator (TKEO) for both raw EMG [15,16] and force signals (Figure 4). The TKEO is a local energy measure for oscillating signals which is proportional to the signal's instantaneous amplitude and frequency [15-19]. In its discrete form, the TKEO (Ψ) value of a signal is given by [15-19]

$$\Psi x(n) = x^2(n) - x(n-1)x(n+1) \tag{1}$$

where x is the EMG or force signal and n is the sample number.

Li et al. [15] and Solnik et al. [16] have shown, using both simulated and real EMG signals, that application of the TKEO effectively suppresses baseline activity where the signals energy is 'low', relative to the time duration of muscle contraction where the signals energy is 'high'. This property of the TKEO is especially useful if the acquired EMG exhibits a low signal to noise ratio or a fluctuating baseline activity, which would bias the calculation of the reference value needed for onset detection when relying on common threshold-based methods [e.g. 20,21]. In this investigation, we used TKEO for detection of onsets because we were concerned with potential false positive onset declaration resulting from heart muscle activity (ECG), a problem that has been observed by several authors in the recording of neck EMG [22,23]. In addition, it was also sometimes difficult to assess the onset of the muscle activity when a given muscle exhibited low levels of activity. Onset of force signals was also processed using the TKEO due to small fluctuations in force baseline activity resulting from small head stabilization movements prior to excursions.

Subsequent to TKEO calculations, EMG and force signals were full wave rectified and low pass filtered using a 2nd order, zero phase shift, butterworth filter with a cutoff frequency of 50 Hz [21]. EMG and force onset thresholds were then set as the instant the signal exceeded 13 standard deviations above baseline levels for a period of 20 ms. All force and EMG onsets were verified visually by a single examiner on the original force and full wave rectified EMG signals. Muscle activation onsets were then calculated as the difference between the onset of EMG and the onset of force (units ms). Negative values indicate that EMG preceded force onset, while positive values indicate that force onset preceded EMG activity onset.

Figure 4.2: Process of EMG (a) and force (b) onset detection: 1a,b) Raw EMG and force signals. 2a,b) Teager Kaiser Energy Operator (TKEO) output for EMG and force signals. 3a,b) Full wave rectified and filtered TKEO outputs. Onset was detected at this stage using preset thresholds. 4a,b) Full-wave rectified EMG and raw force signals used to ascertain onset instances.
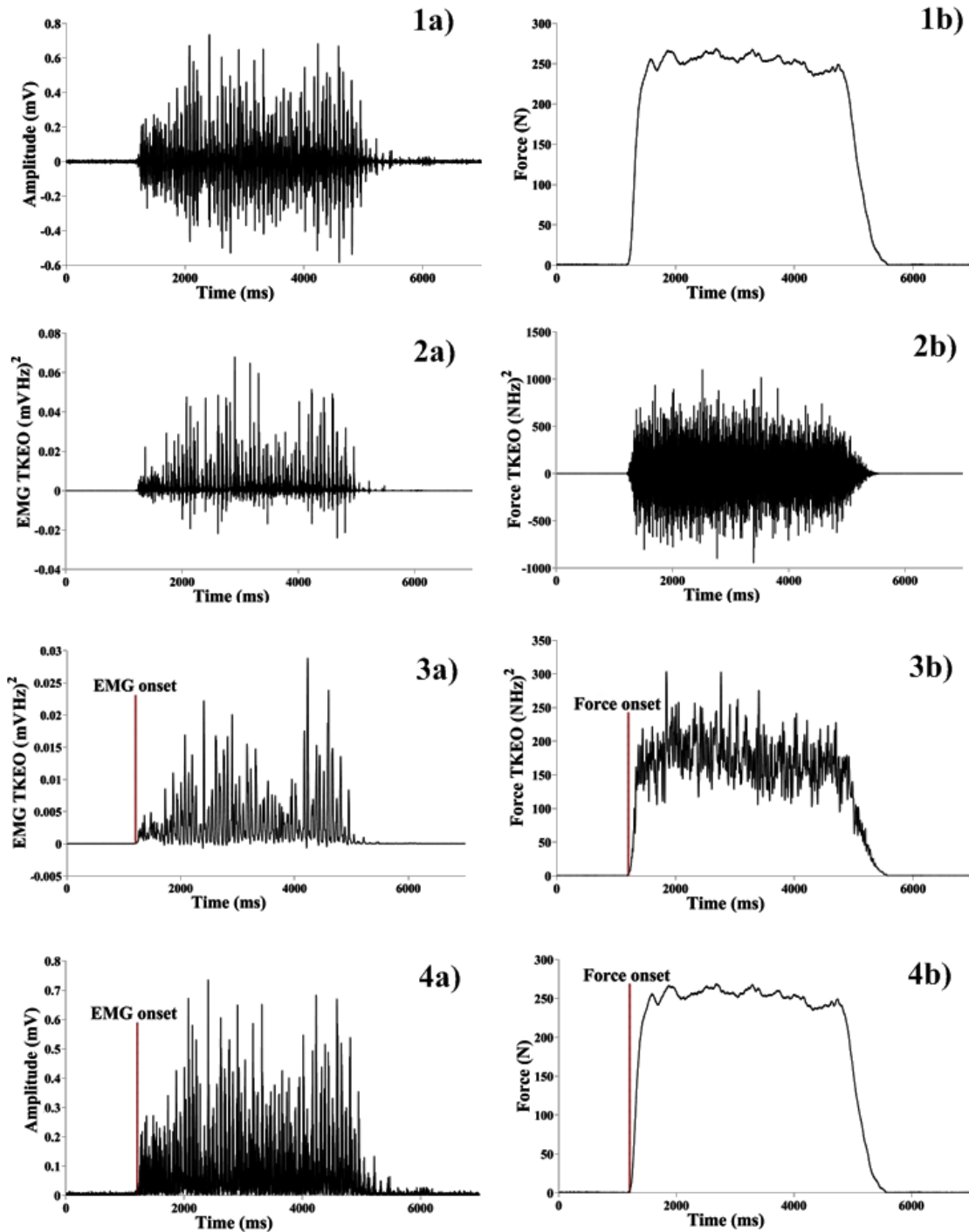
4.6 Statistical analysis

All trials were reviewed, and the 3 trials with the highest rate of force development (e.g., the maximal value of the slope of the force-time curve) were selected. The average muscle activation onset values from these trials were used in the following analysis.

Normality of distribution for muscle activation onset values was assessed for all testing directions by visual inspection of histogram plots and by Shapiro-Wilks normality tests ($p \leq 05$): all data met the normality requirements for parametric statistics. The difference in average scores between testing sessions (i.e. Day 2 score - Day 1 score) were computed along with the corresponding 95% confidence intervals to identify systematic bias [24]. The standard error of measurement (SEM) was used to determine measurement precision. SEM was calculated by dividing the standard deviation of the difference scores by the square root of two [24]. The smallest detectable difference value (SDD), used to determine the smallest change necessary for declaration of statistically significant differences between measurements from the two testing sessions, was calculated by multiplying the SEM by 2.77 [24], and as such utilized a 95% confidence level.

Retest correlation was assessed using intraclass correlation coefficients (ICCs), using the 3, k model [25]. This decision to use this type of model was done after examination of the confidence interval range of the between sessions difference scores, as recommended by Wier [26].

4.7 Results

Table 5 summarizes the results obtained for the muscle activation onsets. Small differences in mean scores of less than 4.0 ms are evident between the two testing sessions across all muscles and testing directions. The corresponding 95% confidence intervals for all comparisons overlap zero, indicating that differences between test sessions are not statistically

significant. Efforts exerted in extension were the most precise, with SEMs ranging from 2.5 ms to 4.0 ms, and corresponding SDD values ranging from 7.3 ms to 11.7 ms. ICC scores for extension ranged between 0.52 and 0.79.

Lateral bending efforts to both sides as well as flexion efforts resulted in slightly higher SEM scores, ranging between 3.4 ms and 4.8 ms. Naturally, the corresponding SDD values were larger for these testing directions, ranging from 9.9 ms to 14.0 ms. ICC scores for these directions are low to moderate, ranging between 0.23 and 0.71. Of note is that some corresponding confidence intervals were negative (e.g. right SpL, right SCM and left SCM in flexion).

Protraction elicited poor reliability for muscle activation onsets, with SEM scores between 8.6 ms and 12.0 ms, and SDD values between 25.4 ms and 35.5 ms. ICC were low to moderate, ranging from 0.31 to 0.77, with some confidence intervals values being negative.

Table 4.1: Muscle activation onsets reliablity indices

	Muscle	D1 Mean (SD)	D2 Mean (SD)	Diff (95% CI)	SEM (95% CI)	ICC$_{3,3}$ (95% CI)	SDD
Ext	R SpL	-31.0 (4.3)	-32.8 (6.8)	-1.8 (-3.9 to 0.4)	3.4 (2.6 to 4.8)	0.79 (0.49 to 0.91)	9.9
	R TRP	-37.4 (3.8)	-38.7 (2.7)	-1.3(-2.9 to 0.3)	2.5 (1.9 to 3.6)	0.59 (0.01 to 0.83)	7.3
	R SCM	3.3 (3.7)	3.0 (3.3)	-0.3 (-2.1 to 1.5)	2.8 (2.2 to 4.1)	0.52 (-0.19 to 0.80)	8.3
	L SpL	-33.1 (5.8)	-35.5 (6.1)	-2.4 (-4.9 to 0.2)	4.0 (3.0 to 5.7)	0.71 (0.29 to 0.88)	11.7
	L TRP	-38.3 (4.5)	-39.4 (4.5)	-1.1 (-3.0 to 0.9)	3.0 (2.3 to 4.3)	0.71 (0.29 to 0.88)	8.9
	L SCM	2.2 (5.5)	1.4 (5.0)	0.8 (-2.9 to 1.3)	3.3 (2.5 to 4.8)	0.74 (0.38 to 0.89)	9.8
Flex	R SpL	-32.8 (4.8)	-32.3 (4.2)	0.5 (-2.2 to 3.2)	4.2 (3.2 to 6.0)	0.23 (-0.88 to 0.68)	12.3
	R TRP	12.0(6.4)	13.9 (5.9)	1.8 (-0.9 to 4.6)	4.2 (5.2 to 6.1)	0.69 (0.25 to 0.87)	12.4
	R SCM	-70.6 (4.9)	-72.0 (4.4)	-1.5 (-4.4 to 1.1)	4.0 (3.0 to 5.7)	0.41(-0.44 to 0.76)	11.7
	L SpL	-30.2 (4.6)	-30.8 (4.8)	-0.6 (-2.7 to 1.6)	3.4 (2.6 to 4.9)	0.66 (0.17 to 0.86)	9.9
	L TRP	9.6 (4.4)	8.4 (5.3)	-1.2 (-3.5 to 1.1)	3.6 (2.7 to 5.1)	0.64 (0.11 to 0.85)	10.5
	L SCM	-69.8 (5.4)	-69.5 (5.0)	0.3 (-2.4 to 3.0)	4.2 (3.2 to 6.0)	0.52 (-0.17 to 0.80)	12.3
Prot	R SpL	-27.5 (9.9)	-27.9 (11.8)	-0.4 (-6.0 to 5.1)	8.6 (6.6 to 12.5)	0.54 (-0.12 to 0.86)	25.4
	R TRP	17.1 (10.8)	20.1 (12.5)	3.0 (-2.9 to 8.9)	9.2 (7.0 to 13.2)	0.77 (0.43 to 0.91)	27.0
	R SCM	-67.8 (13.9)	-66.4 (16.2)	1.4 (-6.3 to 9.2)	12.0 (9.2 to 17.4)	0.54 (-0.14 to 0.81)	35.5
	L SpL	-26.7 (12.3)	-27.9 (18.1)	-1.2 (-7.7 to 5.3)	10.1 (7.7 to 14.6)	0.73 (0.34 to 0.89	29.8
	L TRP	18.2 (10.4)	18.0 (11.9)	-0.2 (-6.7 to 6.2)	10.0 (7.7 to 14.4)	0.32 (-0.66 to 0.72)	29.5
	L SCM	-68.1 (11.0)	-64.2 (13.6)	3.9 (-3.3 to 11.2)	11.3 (8.6 to 16.3)	0.31 (-0.68 to 0.72)	33.2
LLB	R SpL	18.8 (4.1)	19.5 (5.5)	0.7 (-1.9 to 3.4)	4.1 (3.1 to 5.9)	0.46 (-0.33 to 0.78)	12.0
	R TRP	-20.2 (4.9)	-19.8 (5.3)	0.4 (-2.2 to 2.9)	3.9 (3.0 to 5.6)	0.59 (-0.01 to 0.83)	11.5
	R SCM	0.7 (5.6)	1.3 (4.3)	0.5 (-2.1 to 3.1)	4.0 (3.1 to 5.8)	0.51 (-0.19 to 0.80)	11.9
	L SpL	-63.0 (6.0)	-62.2 (5.8)	0.8 (-2.0 to 3.6)	4.4 (3.4 to 6.3)	0.61 (0.05 to 0.84)	12.9
	L TRP	-43.8 (5.8)	-42.7 (4.3)	1.1 (-1.2 to 3.5)	3.7 (2.8 to 5.3)	0.64 (0.11 to 0.85)	10.9
	L SCM	-58.8 (6.0)	-56.4 (6.0)	2.5 (-0.2 to 5.1)	4.2 (3.2 to 6.0)	0.68 (0.22 to 0.87)	12.3
RLB	R SpL	-61.4 (5.2)	-59.6 (3.3)	1.8 (-0.6 to 4.2)	3.8 (2.9 to 5.5)	0.39 (-0.48 to 0.75)	11.1
	R TRP	-38.0 (4.6)	-39.7(5.3)	-1.7 (-4.0 to 0.6)	3.5 (2.7 to 5.1)	0.65 (0.15 to 0.86)	10.4
	R SCM	-62.9 (6.5)	-63.2 (6.3)	-0.3 (-3.4 to 2.8)	4.8 (3.6 to 6.9)	0.61 (0.05 to 0.84)	14.0
	L SpL	28.9 (6.7	29.8 (5.6)	1.0 (-1.7 to 3.6)	4.1 (3.2 to 6.0)	0.71 (0.28 to 0.89)	12.2
	L TRP	-18.5 (6.9)	-16.9 (6.1)	1.6 (-1.2 to 4.5)	4.4 (3.4 to 6.4)	0.70 (0.27 to 0.88)	13.0
	L SCM	2.3 (5.2)	1.4 (5.0)	-0.9 (-3.4 to 1.5)	3.8 (2.9 to 5.5)	0.60 (0.02 to 0.83)	11.3

Ext = Extension, **Flex** = Flexion, **Prot** = Protraction, **LLB** = Left lateral bending, **RLB** = Right lateral bending**, SpL** = Splenius Capitis, **TRP** = Upper Trapezius, **SCM** = Sternocleidomastoid. **D1** = Day 1, **D2** = Day 2, **SD** = Standard deviation, **Difference** = Difference between means, **CI** = Confidence intervals, **ICC** = Intraclass correlation coefficient, **SDD** = Smallest detectable difference **(ms)**

4.8 Discussion

Measurement of neck neuromuscular functions using EMG could provide valuable information to evaluate the effects of interventional programs as well as possibly provide objective measures about readiness to return to play following a concussion injury. However, establishment of the between-day reliability of relevant measures is a prerequisite if they are to be used as part of assessment procedures.

With regards to the results obtained in this study, it should be first noted that no significant differences in muscle activation onset values were observed between the two testing sessions across all muscles and directions. It has been previously recommended with respect to neck muscular testing that a familiarization session be performed, as participants are usually not accustomed to performance of maximal efforts [27,28]. However, our sample consisted of participants who were highly physically active, and we postulated that a well designed pre-testing familiarization and warm-up routine would suffice in order to eliminate any learning effect or apprehension in eliciting such maximal efforts. This is an important finding, as performing a familiarization session necessarily entails the allocation of resources and extra participant involvement, both of which can be limiting factors from a practical perspective. Even so, in different participant populations, such as those suffering from impairment, we certainly agree that a familiarization session prior to testing may be warranted.

Forming definitive statements regarding the acceptability of muscle onsets' measurement precision was difficult, as only a few researchers have reported upon muscle onset reliability indices [29-33], and even fewer have reported upon the long-term effects of interventional programs on muscle activation onset improvements. [34-36]. None of these studies addressed the activation onsets of the neck muscles. In addition, these studies used dynamic exercises as the primary intervention, and in that respect, their results are not as relevant to the static muscle

contraction mode employed in this study. We have been able to identify only one investigation that used an isometric-based interventional program and measured its influence on the muscle activation onset (electromechanical delay, EMD). Kubo et al [36] report an average decrease in EMD of 15.3 ms following a 12 week training program of the knee extensors. Given that the SDD values obtained in our investigation for all efforts, barring protraction, fall well within this improvement, and that large improvements may be expected from those untrained in specific muscular conditioning [37], the measurement precision of activation onsets is deemed to be acceptable for future clinical and research purposes, except in the direction of protraction. While the reasons for the poor reliability of muscle activation onsets measured in protraction are not completely clear, it is possible that providing more practice in this specific direction during the pre-testing procedures could improve outcomes.

Whilst the degree of precision of the muscle activation onsets seems to be acceptable for the majority of movement directions, the ICC values obtained are generally considered low to moderate. ICCs are usually reported as part of reliability assessment because they convey whether different individuals may be distinguishable from one another [11,26,38,39]. However, amongst the shortcomings of ICC is the fact that, if the participants' score range is homogenous, then the magnitude of between-subject variability may closely resemble the magnitude of the within-subject variability, ultimately yielding a low or even negative ICC ratio [11,24,38,39]. In this study, examination of the muscle activation onsets standard deviation values suggests that subjects' score range was indeed narrow, thus being the determinant factor for the low ICC scores obtained. However, given the precision of measurements, the onset values obtained may be used as reference values in subsequent investigations employed with subjects exhibiting similar physical characteristics [11,26,38,39].

4.9 Conclusion

The results of this investigation suggest that activation onsets of neck muscles during isometric MVEs can be measured with an acceptable level of precision. In addition, the lack of difference scores between testing sessions suggests that in highly trained participants, a well-designed pre-testing protocol may suffice in order to eliminate apprehension or learning effects. Based on these results, we plan to utilize the muscle activation onsets as an outcome measure in subsequent investigations involving conditioning of the neck musculature for sport participation, as well exploring whether this measure might be used to assess readiness to return-to-play following injury.

4.10 References

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Chapter 5
General Discussion

5.1 Summary of findings

The purpose of this thesis was to assess the between- day reliability of selected time-dependent force indices and the activity onsets of selected superficial cervical muscles during the performance of maximal, isometric exertions in five different directions. Based on the results of this investigation, it can be concluded that all measures, barring the muscle onsets in protraction, can be recorded with a sufficient acceptable precision for future intervention and clinical studies, and that force-time based variables distinguish different participants from one another in a sample that is highly homogenous in terms of age and physical activity levels.

The decision regarding the acceptability of the precision level obtained in this study for both force- time based variables and muscle onsets was primarily based on the documented change in these variables as a result of an intervention. Meaning, if changes in performance documented in the literature exceeded the range of scores covered by the SEM and SDD for a certain measure, than the precision was declared to be adequate. However, the assessment of precision may have greatly been enhanced if data obtained in this investigation would have served as inputs for the head and neck model of Viano et al. (3), which in turn, would have resulted in the evaluation of neck stiffness outputs on head acceleration values. Unfortunately, this was not done primarily because the implementation of model is not a straight forward matter. However, given that a model of the head and neck is currently developed at Queen's University, the assessment of precision using the model may be done in the near future.

The design of the testing apparatus, as well as the procedures and results of this investigation have addressed several of the points made by Dvir and Prushansky [1] and Sommerich et al. [2] regarding the current practices and knowledge of CS and neck muscle EMG measurements. Specifically, novel information regarding the time-dependent force- generating capabilities of the cervical musculature is presented. These include: the rate of force development and the activity onset of different muscles in different isometric movement efforts. In addition, this investigation demonstrated that maximal efforts in healthy individuals can be measured safely after proper screening and well- designed familiarization and practice sessions. Most importantly from a practical viewpoint, it is noteworthy that learning effects can be diminished in subjects unaccustomed to the types of efforts performed in this investigation. As such, it is hoped that the results of this investigation may provide a sound basis for future investigations aimed at enhancing cervical musculature capabilities in athletes participating in contact sports. In addition, these data will serve as inputs for the validation of biomechanical models of the head and neck; a model is currently being developed at Queen's University.

5.1.1 Limitations

Several limitations are inherent to the investigation performed. First, measurements were restricted to only 'neutral' head posture. This limitation was necessary because of the overall length of each testing session (approximately 90-120 minutes for the 1st session and 60-75 minutes for the 2nd testing session). The vast majority of these time periods were spent either adjusting the apparatus to optimally fit each subjects' individual dimensions (i.e. load cell height and chair distance from the load cell) or verifying that the subject was in the self-selected head neutral posture recorded prior to being secured to the apparatus. This limitation may be overcome

in future iterations of the testing apparatus by changing using fastening bolts that incorporate levers, similar to those seen in commercial isokinetic dynamometers.

A second limitation of this study is that rotational efforts were not recorded. Current investigative results of concussion injury mechanisms suggest that rotational acceleration plays a large role in determining the occurrence and severity of this type of injury [3]. Therefore, it would seem pertinent that these efforts also be quantified. The current investigation has not done so primarily because the commercial hockey helmet did not allow proper stabilization of the head when performing these types of efforts. A lack of proper head stabilization would surely raise concerns about the validity of EMG recordings. In addition, adding measurements of bilateral rotations would require subjects to perform nearly 30 maximal voluntary exertions, as well as lengthen pre-testing familiarization. Additional testing would possibly result in added muscular fatigue. It should be noted that the stabilization of the head for measurement of rotational efforts is easily attainable using football helmets (Riddell® Pro Evolution), which have been piloted in the current apparatus. Football helmets incorporate inflatable pads that are located to the sides of the mandibles and the skull, and prevent relative motion of the head with respect to the helmet when subjects performed maximal rotation efforts. On the other hand, it should be noted that participants not accustomed to wearing this type of helmet reported severe discomfort, which in turn might have hindered the performance of maximal exertions if it were to be used. However, if the target populations of future investigations are in fact football players, then using their typical type of helmet they regularly don during sport participation would be quite beneficial. On the other hand, if a hockey helmet is to be used in future investigations, than adjustments must should be made to allow support to be given at the sides of the mandibles and skulls so that for valid recordings of rotational efforts can be made.

An additional limitation relates to the fact is that neither type of helmet currently available for interfacing with the testing apparatus allows for measurement of pure extension efforts. This limitation primarily stems from the lack of support provided under the occipital proturbence. The hockey helmet does attempt to provide some support by way of a supporting pad that can be tightened using Velcro® straps. However, while performing pure extension efforts, the tension provided by these straps is inadequate, subsequently resulting in the head sliding underneath the strap. Once again, future design improvements of the helmet should address this issue.

Only maximal isometric efforts were performed in this investigation. If the device is to be used in the evaluation of symptomatic subjects, it would be pertinent to investigate the reliability of sub-maximal exertions, since it is doubtful whether this type of population will be able to exert maximal efforts.

5.1.2 Future Research

The establishment of the between-day reliability of the current testing apparatus and experimental protocol inherently suggests that future investigations should focus on the effects of interventional programs on cervical muscular parameter improvements in the context of concussion prevention. In addition, the device might be also used in order to collect baseline abilities pre-season, followed by a prospective investigation that will explore the role of cervical muscular capabilities and concussion incidence in sports. It would also seem relevant to perform cross sectional studies with symptomatic-, age- and physical activity- matched individuals and compare them with the values obtained in the current investigation in order to explore possible differences in cervical neuromuscular function. In addition, it would be interesting to explore the relative coordination of the neck musculature via EMG amplitude estimates. Finally, it is

important that future research will inherently target adolescents and females, who seem to be at-

risk populations for concussion occurrence in sport settings.

5.1.3 References

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Appendix A

Letter of introduction and consent form

Dear Queen's student,

A group of researchers from across the Province of Ontario are conducting a prospective study of the ice hockey behaviors of youth hockey players. The main purpose of this project is to evaluate the effectiveness of the *Play It Cool* program that was created by Dr. William Montelpare and his research team at Lakehead University to foster safe on-ice behaviors for youth hockey players by providing coaches with educational material. One particular area of high interest for this project is to understand more completely the relationship between neck muscle strength and coordination and a player's ability to control loads that are applied to the upper body and head during on-ice contact. Our research study is related to this specific issue.

To measure the change in the strength and coordination of neck muscles in youth hockey players, 8 to 16 years old, we have developed a measurement jig that is shown below. We are asking you to participate in a pilot study to establish the test-retest reliability of our measurement device and experimental protocol.
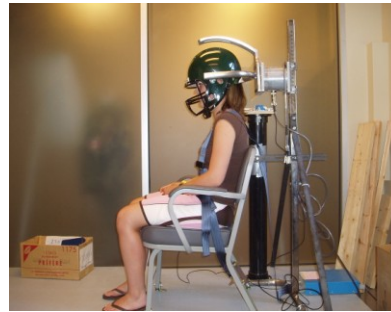
Objectives of the study ➲

To measure the difference in strength and coordination of the muscles of your neck, tested at a 1-week interval.

Methodology ➲

Personal information: To participate in our study, you need to be a male, over the age of 18 years and currently not involved in a neck strengthening program. For your personal safety, we wish to exclude you from the study if you are currently suffering from a neck injury. If you qualify to take part in our study, we will use a code number instead of your name on all paperwork and computer files used in the study. All of this information will be kept in a locked file in a password-protected computer.

The neck muscle strength test: First we will adjust the equipment so that it 'fits' you correctly, allowing you to generate your best strength effort. We will ask you to sit in a chair and we will adjust the head apparatus to align a force gauge to the height of your ears. Our measuring device

uses a football helmet rather than a hockey helmet – the football helmet fits more securely to your head when we test your muscle strength.

The second step is to ask you to perform some neck and upper body stretching exercises. These exercises will be similar to the ones that you perform in hockey practice. The researcher will lead you through these exercises for 3-5 minutes. When you feel that you are warmed up and ready, we will begin with the next step.

The third step is to have you sit in the testing apparatus that was previously fitted to you, and a light-weight harness system will be adjusted to restrain movement of your trunk and shoulders during the testing. We will then tighten the helmet to the force gauge. We will then add some air to the helmet bladders so that it is as snug as possible (same snugness as football players should wear during a game). If it is too snug, please tell us and we will remove some air. Using an electronic pointer, we will then locate specific sites on your head and neck so that we can define your sitting posture. We use this information to calculate the amount of force that your muscles are producing during the testing. During the test, we want to be able to measure how you are coordinating the muscles of your neck to produce the force that we are recording. To do this, we will place small microphones (or electrodes) on specific muscles of your neck and plug them into the recording equipment. We will use adhesive tape to anchor these electrodes in place.

The fourth step will be to ask you to perform a series of 3 maximum force contractions of the muscles of your neck in 6 different directions, for a total of 18 contractions each lasting 3 to 5 seconds. We want to record your best effort and therefore will give you as much time to rest as you need between contractions. We will then repeat these 18 contractions with your head and neck placed in about 20 degrees of flexion.

* The order in which different participants in the study complete these two tasks will be determined by chance.

Risk and benefits of participation ➲

To protect you as much as possible from risks of injury, we want you to way up well and stop at any point if you experience any pain or unusual discomfort. We will also provide 2 minutes of rest between each trial. If you need more rest time, please just let us know. Despite these precautions, it is possible that you could hurt a neck muscle during the test. If you do hurt yourself, please tell us right away. If we are not available, please contact your coach or trainer or personal doctor. If extremely severe, please go to the emergency department and let us know later.

Some individuals may have sensitive skin that is affected by rubbing alcohol or adhesive tape. Normally, this irritation disappears shortly after the tape is removed. If it does not disappear within a couple of hours, please call us or speak to your coach or trainer. Again, you may go for medical help if necessary.

The main benefits of the study are for youth hockey players as we try to find ways to reduce concussions in ice hockey.

Confidentiality ➲

All information obtained during the neck strength and coordination study is strictly confidential and your anonymity will be protected at all times. All data recorded in the computer files will be locked and only researchers will be granted access. In all cases of publications, only summary data are used and this is done in such a way that no individual can be identified.

Voluntary nature of the study ⮑

       As a participant, you are a volunteer who may withdraw from the study at any time without being pressured. You may withdraw during any one of the testing sessions. If you wish, we will remove your data from our records.

       If you would like to talk to the researchers about any aspect of this research before you make a decision about your son's participation, please fell free to contact any of the people listed below:

Dr. Lucie Pelland (School of Rehabilitation Therapy, Queen's University)
Email: lucie.pelland@queensu.ca   Phone: 613-533-3237

Dr. Bill Montelpare (School of Kinesiology, Lakehead University)
Email: wmontelp@lakeheadu.ca   Phone: 807-343-8481

Dr. Albert Clark (Chair, Research Ethics Board, Queen's University)
Email: clarkaf@queensu.ca   Phone: 613-533-6081

We thank you for working with us on this project.

Sincerely,

Lucie C. Pelland, Ph.D.
Assistant Professor, School of Rehabilitation Therapy
Queen's University, Kingston, Ontario, K7L 3N6,
(613) 533-3237
Lucie.Pelland@queensu.ca

PLAY IT COOL PROJECT
Patterns of neck muscle strength and coordination in youth hockey players :
TEST – RETEST RELIABILITY

VOLUNTARY CONSENT TO PARTICIPATE

This page is the ethics consent page and there are two copies; one is for you to keep and the second one is for the researcher. The purpose of this letter is to describe the test-retest reliability study for the neck muscle strength component of the *Play It Cool* research study and invite you to participate. The goal of this study is:

To measure the difference in strength and coordination of the muscles of your neck, tested at a 1-week interval.

By signing this consent form, you realize that you do not waive your legal rights nor release the investigator(s) and sponsors from their legal and professional responsibilities.

What Does My Signature Mean?
When you sign below, you are declaring the following:
I was given a verbal presentation about the above-mentioned research study
I was given this Ethics Consent letter of information to read and keep
I am aware that the purpose of the study is to assess my neck strength and neck muscle activity on three different occasions over the season
I realize that I can withdraw at any time without pressure.
I know that I can contact any of the people identified in the Ethics Consent letter if I have questions, concerns, or complaints
I realize that my data will be kept confidential to researchers only and will not be given to the coach, trainer or other team members.
I will be given a confidential copy of my personal results in comparison to the team average at the end of the season

(Please sign and return this page ONLY to the researchers or person in charge).

_____   _____
Signature of Participant   Date

_____   _____
Signature of Parent / Guardian (as applicable)   Date

_____   _____
Signature of Researcher   Date

PLAY IT COOL PROJECT
Patterns of neck muscle strength and coordination in youth hockey players :
TEST – RETEST RELIABILITY

VOLUNTARY CONSENT TO PARTICIPATE

This page is the ethics consent page and there are two copies; one is for you to keep and the second one is for the researcher. The purpose of this letter is to describe the test-retest reliability study for the neck muscle strength component of the *Play It Cool* research study and invite you to participate. The goal of this study is:

1. To measure the difference in strength and coordination of the muscles of your neck, tested at a 1-week interval.

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I am aware that the purpose of the study is to assess my neck strength and neck muscle activity on three different occasions over the season
I realize that I can withdraw at any time without pressure.
I know that I can contact any of the people identified in the Ethics Consent letter if I have questions, concerns, or complaints
I realize that my data will be kept confidential to researchers only and will not be given to the coach, trainer or other team members.
I will be given a confidential copy of my personal results in comparison to the team average at the end of the season

(Please sign and return this page ONLY to the researchers or person in charge).

_____   _____

Signature of Participant   Date

_____   _____

Signature of Parent / Guardian (as applicable)   Date

_____   _____

Signature of Researcher   Date

Appendix B

Medical Questionnaire [1]

Subject ID code: _____ *Page 1/2*

MEDICAL HISTORY

The information that you provide about your health on this questionnaire will be kept in a secure location with all other information gathered during the data recording sessions. This information will only be available to the investigators of the study and will not be transmitted to any other party under any circumstance.

Have you ever sustained an injury to the head and/or neck? Yes / No
If yes, please describe:

Have you or are currently suffering from head and/or neck and/or shoulder pain? Yes / No
If yes, please describe:

Have you ever sustained an important injury to the upper extremities? Yes/ No
If yes, please describe:

Have you had surgery to the head or neck or shoulder regions? Yes / No
If yes, please describe:

Have you being diagnosed as being at risk for development of any heart disease? Yes / No
If yes, please describe:

Have you ever been affected by the following disorders?

    Joint disorders (including inflammatory diseases) Yes / No
If yes, please describe:

    Visual disorders
    Yes / No
If yes, please describe:

c) Epileptic disorders and/or frequent fainting and/ or dizziness   Yes / No

If yes, please describe (and include information on current medication):

d) Elevated blood pressure levels   Yes / No

If yes, please describe (and include information on current medication):

e) Carotid or coronary artery disease   Yes / No

If yes, please describe (and include information on current medication):

Are you currently taking any medication or other health products?   Yes / No

If yes, please describe:

Do you have any other medical condition that should be mentioned?   Yes / No

If yes, please describe:

Do you currently participate in sports/physical activities?   Yes / No

If yes, please describe:

_____ __   _____
Participant Name (Please Type)   **Investigator Name**

_____   _____
Participant Signature   **Investigator Signature**

_____   _____
Date   **Date**

1. Sommerich CM, Joines SM, Hermans V, Moon S. Use of surface electromyography to estimate neck muscle activity. *J Electromyogr Kinesio*l 2000; 10(6):377-98.

Appendix C
Verification of Load Cell Calibration

The following procedure was employed to verify the accuracy of the factory calibration of the load cell used to record the forces exerted by subjects in this investigation. The load cell used in this investigation as a recording capability of up to 11,111 N along its longitudinal axis (F_Z), and up to 4566 N on both the corresponding orthogonal horizontal axes (F_X and F_Y). The factory calibration procedure uses application of known loads of magnitudes that far exceed those expected to be exerted by subjects in the experiment.

Procedures

The load cell (model MC5-6-2500, AMTI, Watertown, MA, USA) was place on and firmly affixed to a horizontal pedestal, and interfaced with a Modular 600 multi-channel transducer amplifier (±10v peak to peak range, frequency response 0-1kHz, CMRR 110 dB at 60 Hz, input impedance 100MΩ, RDP Group™, Pottstown, PA, USA) and sampled using a 16-bit Analog to Digital (A/D) converter (NI PCI-6036E, range of ±5V) at 2048Hz using custom-software written in Labview™ version 8.6 (National Instruments Inc., Austin, TX, USA). The amplifier was turned on and was left untouched for 20 minutes prior to recordings. Three trials were recorded with the no loads attached to the load cell in order to record baseline offsets. Thereafter, a 70 x 4 x 2 cm wooden strut was attached to the load cell's surface by way of two counter sunk threaded bolts. The position of the load cell was adjusted in order to record all 6 outputs (*Fx, Fy, Fz , Mx, My , Mz*). Known weights (±0.01kg) were hung from the strut using a non-flexible cable at a distance of 0.12m (±0.01m) from load cell's center. The magnitudes of the weights were: 0.9, 1.9, 2.9, 3.9, 4.9, 5.9, 6.9, 7.9 and 8.9 kg. The weights were hung from the strut in a sequential ascending order and were allowed 45-60 seconds to reach a resting position

before the recording of a five second static trial. A total of three trials were recorded for each weight iteration.

Analysis and Results

Load cell voltage outputs were converted to Newtons according to the calibration values provided by the manufacturer. All channels were zero offset prior to calculation of the mean value of each channel for each trial. Subsequently, a grand average was computed for each weight iteration in each of the load cell positions. The accuracy of the load cell outputs was assessed by computing the root mean square error (RMSE) for all channels. In addition, output linear responses were assessed by calculating least square parameter estimates, and by subsequent evaluation the coefficient of determination (R^2). Table 6 summarizes the results. For all force channels, the RMSE was below 1N, and for all moment channels, the RMSE was below or equal to 0.1 Nm. Expressed as a percentage, RMSE for forces ranged between 1.0 to 1.25%, and for moment channels between 1.4 to 1.9%. The coefficient of determination was above 0.99 for all force channels and moments, indicating a linear response within the measurement range (Figure C).

Table C1: Forces and moments channel outputs for known applied loads

	Force Channels				Moment Channels		
Load applied (N)	Fx Output	Fy Output	Fz Output	Load applied (Nm)	Mx Output	My Output	Mz Output
0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00
8.83	9.01	8.65	8.87	1.06	1.11	1.02	1.07
18.64	18.48	18.37	18.76	2.24	2.24	2.26	2.19
28.45	28.90	28.47	29.19	3.41	3.58	3.49	3.46
38.26	39.03	38.67	38.99	4.59	4.51	4.54	4.61
48.07	48.76	48.32	48.76	5.77	5.72	5.73	5.79
57.88	58.12	58.54	57.99	6.95	7.03	6.89	7.01
67.69	67.12	67.69	68.23	8.12	8.02	8.07	8.21
77.50	78.26	78.11	78.31	9.30	9.37	9.41	9.26
87.31	87.76	88.17	87.96	10.48	10.52	10.22	10.66
RMSE (N)	**0.50**	**0.44**	**0.54**	**RMSE (Nm)**	**0.08**	**0.10**	**0.07**
RMSE (%)	**1.15**	**1.01**	**1.25**	**RMSE (%)**	**1.53**	**1.87**	**1.38**

Figure C1: Load cell outputs to known applied load. Fx, Fy and Fz correspond to the force channels, while Mx, My and Mz correspond to the moment channels.

84
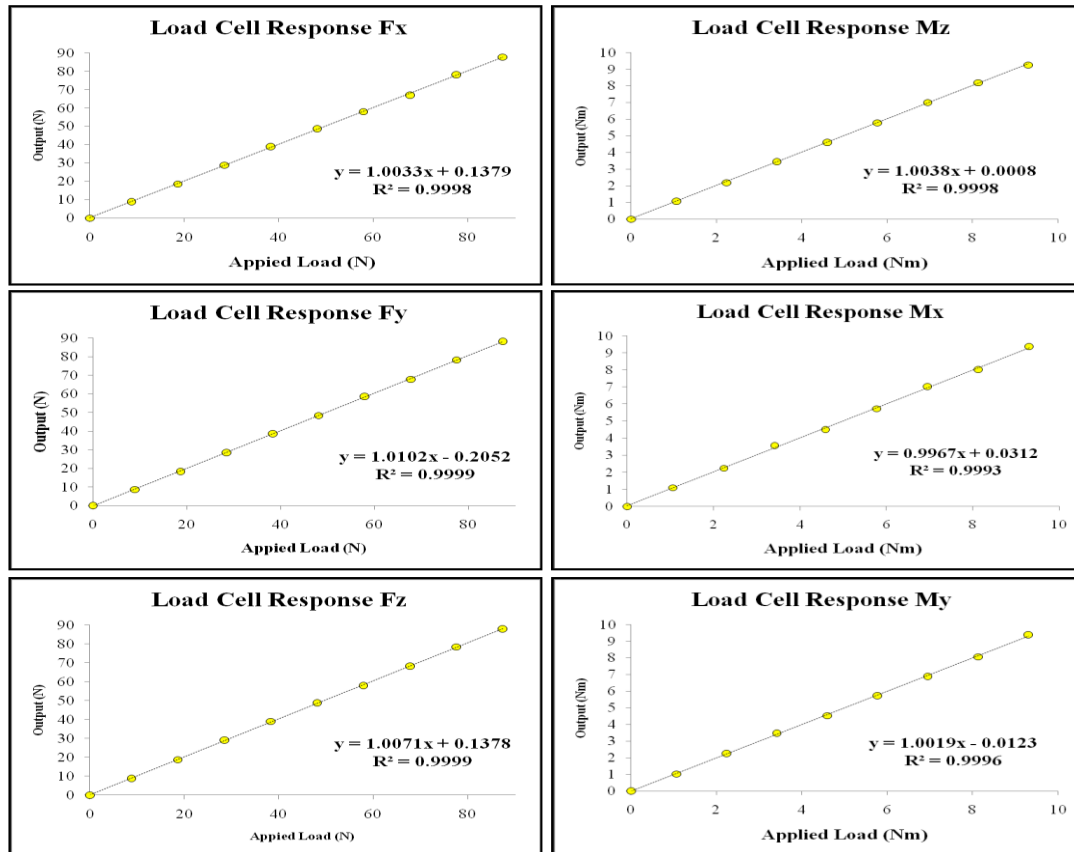
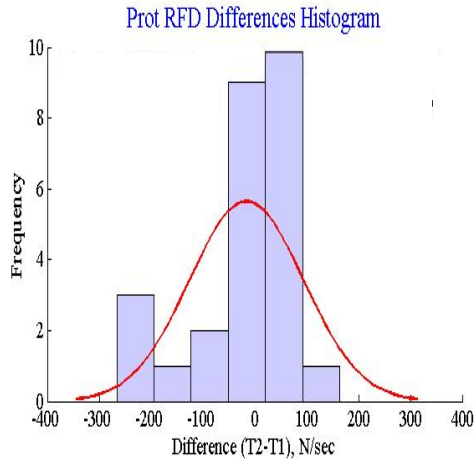## Appendix D
Data Inspection Procedures

The following illustrations (Figures D1 and D2) are of the data inspection procedures that took place prior to performing any statistical analysis. The top left graph in each illustration is a frequency histogram of the difference scores between the means of the two testing sessions. Standardized skewness and kurtosis values were calculated, and a Shapiro-Wilks test was conducted (alpha set at 0.05). The top right graph is a scatter plot of the absolute differences between test 1 and test 2 vs. the mean of the two testing sessions. This scatter plot was used to visually assess the presence of heteroscedasticity (e.g. an increase in the difference between the two test sessions as a function of strength) according to the recommendations of Atkinson and Neville [1]. The assessment of heteroscedasticity was also aided by calculation of the Pearson product moment correlation coefficient (r), which enabled identification of any linear trends pre and post log transformation of the data set.

The example given in the illustrations are for rate of force development values obtained for protraction effort. The raw data visually exhibits some heteroscedasticity, with a statistically significant Pearson correlation value of 0.58. Further, the data is not normally distributed, as evident from the histogram plots, the results of the Shapiro-Wilks test, and the standardized skewness and kurtosis scores. Following log transformation, Pearson's r is reduced to 0.27 and is not statistically significant. Visually, the difference scores are more uniformly spread. The normality of distribution is also improved visually, which consequently leads to a reduced standardized Z skewness score. The Z kurtosis scores was slightly increased, but is well within the normal range. Finally, the Shapiro-Wilks test results are not statistically significant.
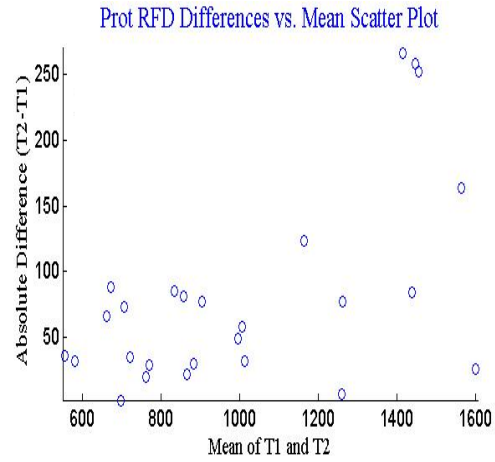
Reference: 1. Atkinson G, Nevill AM. Statistical methods for assessing measurement error (reliability) in variables relevant to sports medicine. *Sports Med*. 1998; 26 (4): 217-38.

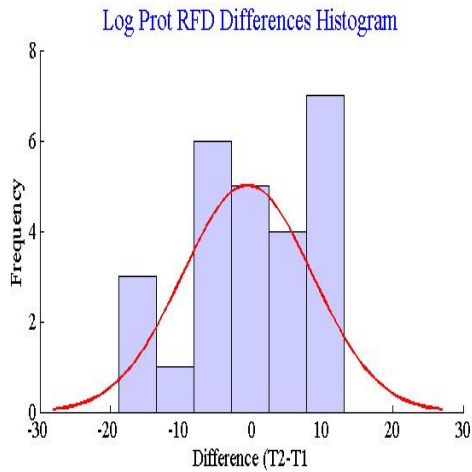**Figure D1: Data Inspection Procedures Pre (1) and Post (2) Log Transformation**

1)

2)
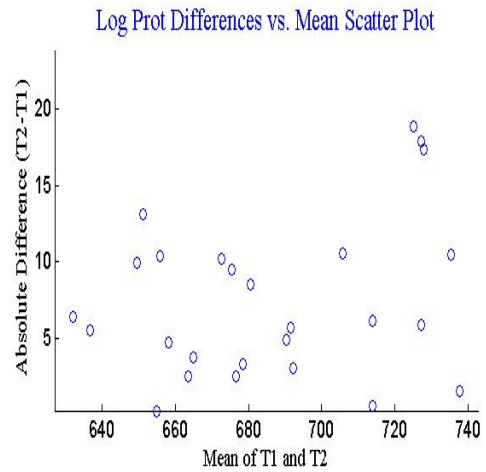
Appendix E

Reliability and accuracy of Isotrak digitized coordinates

The following procedure was completed to quantify the accuracy and between-day reliability of the 3D coordinates obtained using the 3-Space Isotrak digitizer system (Polhemus Navigation Sciences, Colchester, VT, USA). This was done as it suspected that the 3Space Isotrak™ may yield erroneous digitized coordinate values as a result of presence of metallic objects and computer screens within the immediate vicinity of both transmitter and receiver.

A 12cm x 15cm wooden board containing 24 holes spaced at 2.5cm of each other was used in this experiment. The board was attached to the testing apparatus using the same methods used to append the helmet, and leveled with respect to the horizontal. The position of the 24 holes was then digitized by a single examiner with the pen stylus housing the receiver. The pen stylus was held vertically and in the same Isotrak source quadrant. Digitization was done at four different vertical distances (0cm through 20cm), where the vertical distance is expressed relative to the first digitization vertical position. The number of threads that corresponded to each vertical distance were recorded for use in the second testing session.

The second testing session was done seven days later. During that period, the board remained attached to the testing apparatus, such that its distance relative to the Isotrak source box was un-altered. The vertical positions of the board were adjusted to correspond to those in the first testing session by placing the top of the board exactly at the same thread-count level recorded during the first testing session. During both testing sessions, all electrical equipment was turned on (e.g. load cell amplifier, EMG amplifier, visual feedback screen).

Analysis

As the true location of the points relative to the Isotrak source box was unknown, the position of all holes was expressed relative to the position of the rightmost hole in the 0cm vertical position (0,0,0 position). For the assessment of between-day systematic error, differences between the 1st and 2nd testing session average scores were calculated assessed with a paired t-test with alpha preset at 0.05. The standard error of measurement (SEM) was used to assess the precision of measurements between the two testing sessions, and was calculated by dividing the standard deviation of the 1st and 2nd test sessions difference scores by the square root of two. The maintenance of rank order of the digitized points (in terms of distance from the origin) was assessed by the Pearson product moment correlation. Accuracy of measurements was assessed by plotting the average difference scores between the 'true' board dimensions and those obtained by digitization.
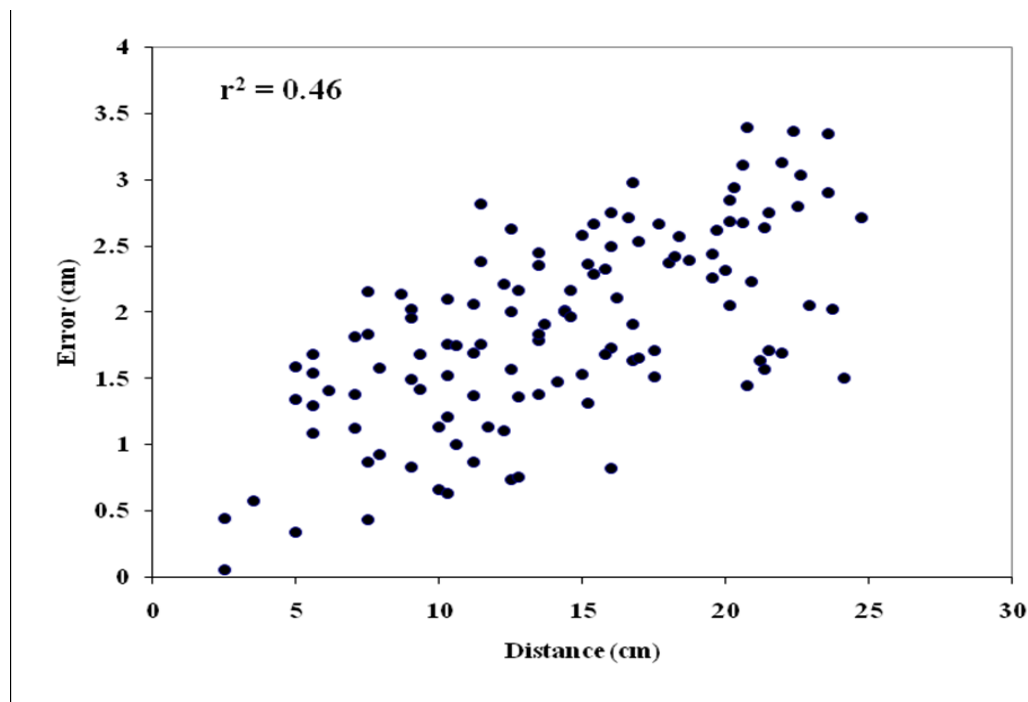
Results

Average digitized distances recorded during the 1st and 2nd testing sessions were (mean (SD)) 15.91cm (6.18) and 15.87cm (6.12), respectively. The results of the paired t-test indicate that the difference between testing sessions was not statistically significant (t (118) = 1.80, p = 0.07). The SEM value was small (0.19 cm), indicating a high degree of precision in between- day measurements. Relative reliability, or the maintenance of rank order among digitized points, was very good, with *r* > 0.99.

As no differences were evident in scores obtained from the two testing sessions, the assessment of Isotrak accuracy was done only on the 1st test session results. Figure E depicts the error in digitized coordinates obtained in the 1st test session as a function of the distance from the origin point. A clear positive trend is noticeable; the error magnitude increases the farther the

digitization point is from the source. Evaluation of the coefficient of determination (r^2) shows that 46% of the error variance can be attributed to the digitized distance from the source. The rest of the variance can be speculated to be a result of electro-magnetic interference, inaccuracies in the distance between the points on the wooden board, and small examiner-induced errors in placement of the pen stylus.

Figure E1: Isotrak digitizing error as a function of distance from the source

Conclusion

The use of the Isotrak digitizer system in this investigation is intended for quantifying the head and neck positions of the participants in a test-retest scheme. The results indicate that the Isotrak can be used for this purpose, as it shows very good between-day reliability in both absolute and relative terms. However, the accuracy of the Isotrak proved to be poor. Thus, the head and neck angles presented in Appendix F are not reflective of the true position of the participants' head and neck during performance of trials.

Appendix F

Measurements of Head and Neck Position

Subject's self selected neutral head position within the testing apparatus was established by digitizing the 3D coordinates of seven anatomical points using a 3Space Isotrak digitizer system (Polhemus Navigation Sciences, Colchester, VT, USA). Head position was defined using the following anatomical points: left and right ear tragii, left and right lowermost border of the orbit, C2 spinous process, C7 spinous process and the shallow-most portion of the sternal notch. Subsequently, the following angles were calculated in order to quantify the head posture in the sagittal plane [1]: a) Head Angle (HA), defined as the angle between the horizontal and the line segment originating from the midpoint of the ear tragii line segment to the midpoint of the orbit line segment, and b) Neck angle (NA),defined as the angle between horizontal and the line segment connecting the C7 and the midpoint of the ear tragii line segment. Table 7 lists the individual HA and NA obtained in each of the testing sessions. The criteria for set for subjects' head position to be considered equivalent to that recorded during the first test session was that all angles were within 2 degrees of each other. Note that the results of 28 subjects are listed; the initial 26 subjects are those listed for study 1. A sub sample of these subjects (1-19) and the last 2 subjects are the ones for whom EMG results are reported in study 2.

Reference:

1. Joines SM, Sommerich, CM, Mirka,GA, Wilson, JR, Moon, SD. Low-level exertions of neck musculature: a study of research methods. *J Electromyogr Kinesiol*. 2006; 16 (5), 485 -97.

**Table F1: Individual Head and neck angles recorded in each test session**

Subject Number	Test 1 HA (degrees)	Test 2 HA (degrees)	Difference (degrees)	Test 1 NA (degrees)	Test 2 NA (degrees)	Difference (degrees)
1	34.6	34.1	0.5	139.5	137.6	-1.9
2	25.6	23.8	1.8	125.2	123.4	-1.8
3	19.8	21.0	-1.2	124.5	123.1	-1.4
4	27.9	26.1	1.8	134.4	136.0	1.6
5	33.6	31.8	1.8	139.1	140.6	1.5
6	23.2	21.7	1.5	113.5	112.8	-0.7
7	31.1	32.5	-1.4	134.8	135.4	0.6
8	42.0	42.3	-0.3	117.3	118.9	1.6
9	37.5	38.1	-0.6	144.4	144.1	-0.3
10	34.4	34.2	0.2	130.6	132.2	1.6
11	17.3	18.9	-1.6	125.5	127.4	1.9
12	26.0	25.5	0.5	133.6	131.8	-1.8
13	20.9	21.1	-0.2	141.4	142.3	0.9
14	24.4	26.2	-1.8	122.4	123.7	1.3
15	20.9	22.1	-1.2	151.2	151.4	0.2
16	35.2	34.4	0.8	133.7	132.1	-1.6
17	20.8	19.1	1.7	117.7	119.5	1.8
18	18.8	19.1	-0.3	139.8	137.9	-1.9
19	21.7	23.1	-1.4	126.8	127.7	0.9
20	27.1	25.9	1.2	112.5	114.3	1.8
21	32.6	32.1	0.5	140.1	141.5	1.4
22	29.9	28.4	1.5	131.1	129.8	-1.3
23	25.2	25.7	-0.5	120.1	118.8	-1.3
24	36.6	37.5	-0.9	148.8	147.8	-1.0
25	16.3	17.8	-1.5	136.9	137.6	0.7
26	19.3	20.1	-0.8	125.5	125.8	0.3
27	25.7	27.5	-1.8	134.2	135.9	1.7
28	36.8	35.2	1.6	122.1	123.4	1.3
Mean	**27.3**	**27.3**		**131.0**	**131.2**	
SD	**7.1**	**6.8**		**10.2**	**10.0**	

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Appendix G

Computer Programs

Several custom data acquiring and analysis programs were written for this investigation. These programs were primarily constructed in Labview (version 8.6; National Instruments Inc., Austin, TX, USA), though the analysis section relied heavily on Matlab scripts (version 7.5; MathWorks Inc., Natick, MA, USA) that were incorporated into the Labview environment. Figure G1 presents the operator front panel of the data acquiring program. This program contains several features for the use of future planned investigations: Features include the ability to record sub-maximal efforts at a preset level of maximal force with an adjustable error range as defined by the operator. Visual and audible cues are generated to indicate to both the operator and participant that the sub-maximal forces are within the defined range.

Figure G1: Force and EMG data acquiring program.
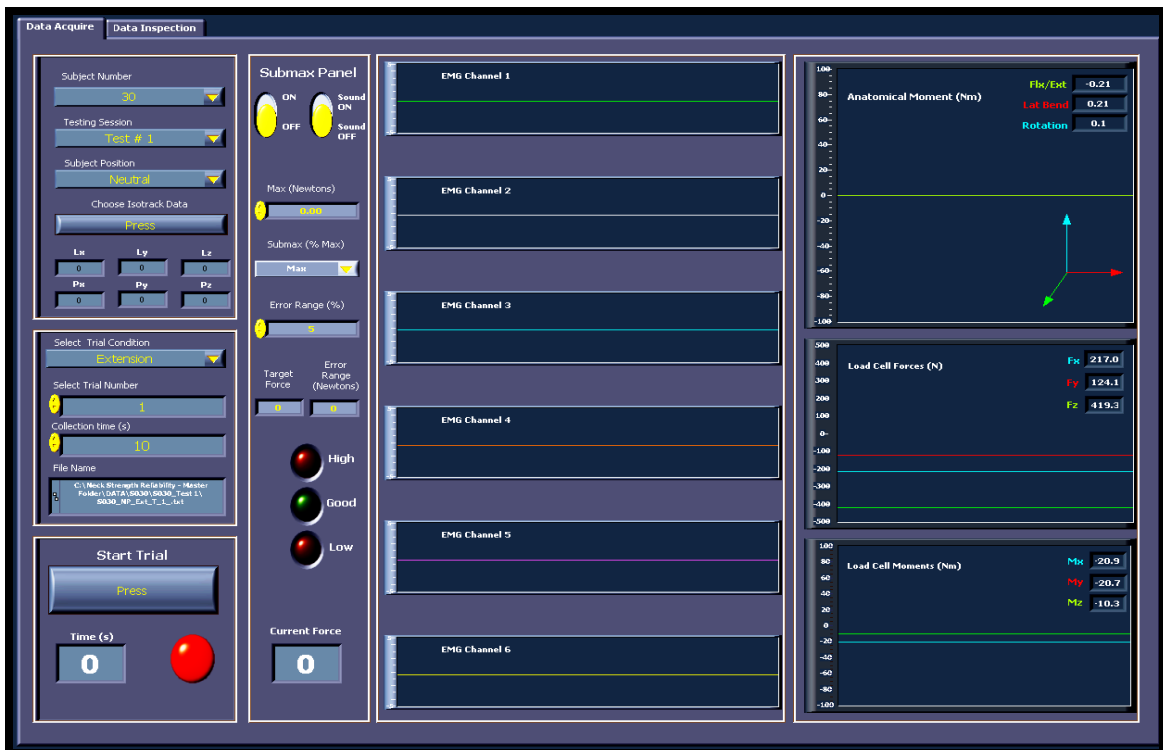
The data inspection part of the acquiring program is shown if Figure G2. This feature was used for inspection of the raw force and EMG trials during the subject rest period.

Figure G2: Data inspection part of the data acquiring program

The real-time visual feedback provided to the participants during the performance of the trials is shown in Figure G3. The three-dimensional "smiley face" moves along the principal axes in correspondence to the load cell outputs recorded during the trial. The bar graphs indicate the force value exerted by the participant. The red and green indicators are the visual feedback provided to the participant during the performance of sub-maximal trials. If the participants force range is within that preset by the examiner, then the green indicator is lit. If the force range is outside this range, the proper red indicator is lit. The design of the visual feedback differs from

the traditionally supplied force-time curve because it was designed for children, which might find real-time interpretation of the force-time curve difficult.

Figure G3: Real-time participant visual feedback screen.

The data analysis program is shown in Figure G4. Some features of this program include: the ability to load data files from both the 1st and 2nd testing sessions in one instant, instantaneous screen updates based on the user- selected analysis (i.e., enabling, for example, assessment of the influence of different filter orders and cut-off frequencies), and manual adjustment of force and EMG onsets if not detected properly by the algorithm utilized. The left-bottom window is the force-time curve with corresponding outputs as elaborated upon in chapter 3. The right-bottom indicators are the outputs of the EMG signals, as elaborated upon in chapter 4. Other analysis features include EMG frequency analysis and APDF. Finally, a computerized visual-analog scale was constructed to record subject comfort following each testing session

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(Figure G5). This data were collected for internal purposes concerned with improvements of the
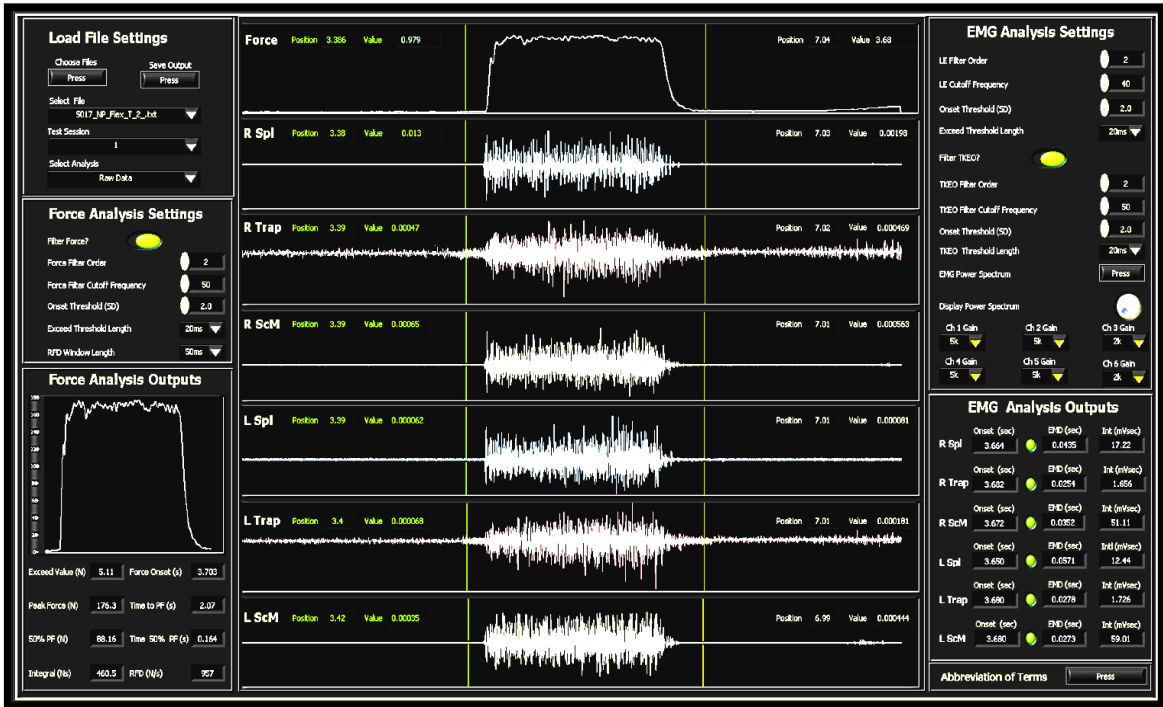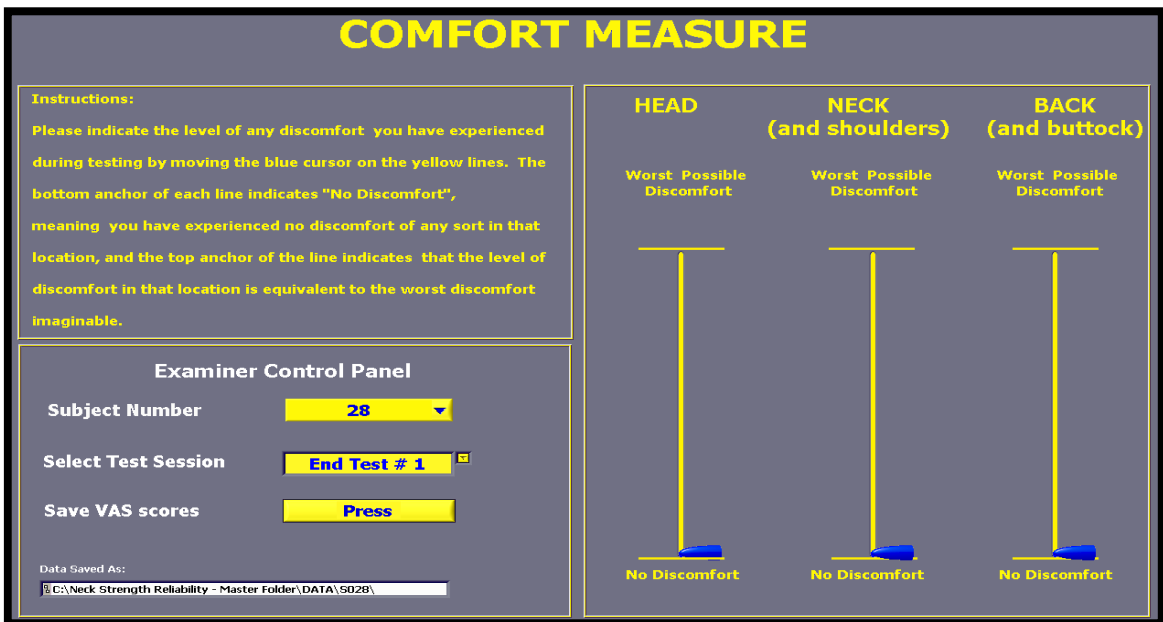
testing apparatus.

Figure G4: Data Analysis Program

Figure G5: Computerized visual analog scale (VAS)

Appendix H

Testing Apparatus Dimensions
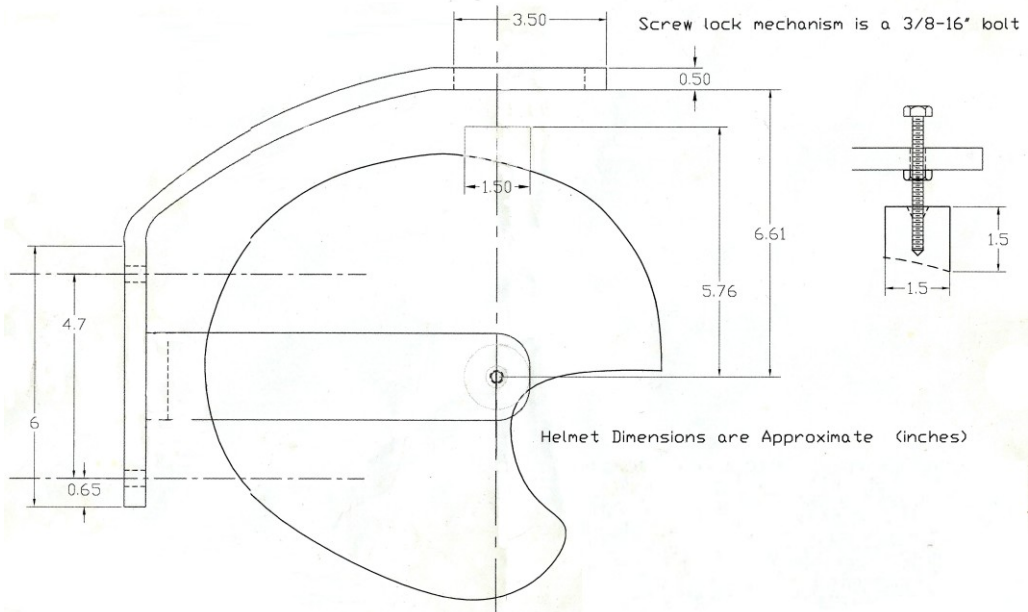
Figure H1: Helmet Restraint Structure (Side)

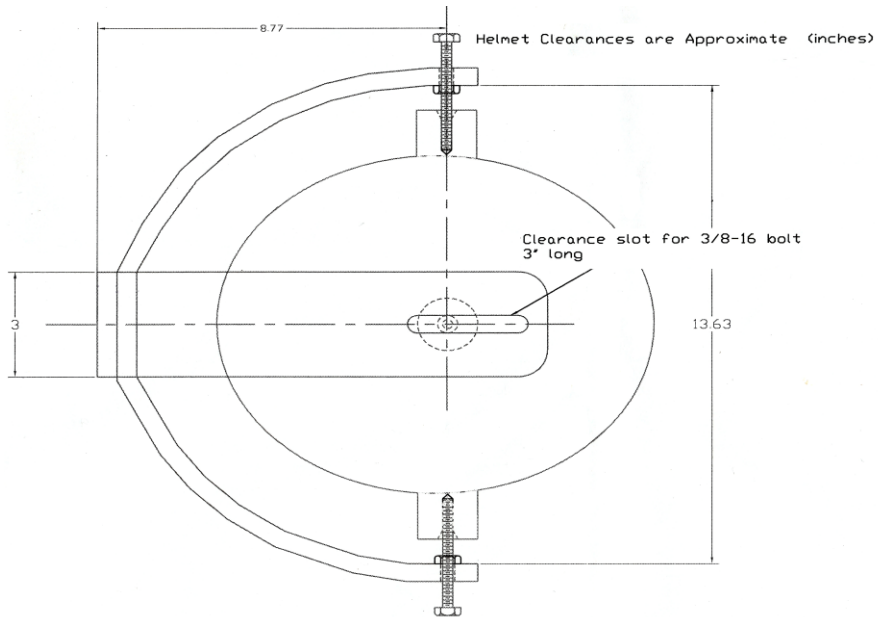**Figure H2: Helmet Restraint Structure (Top)**

Figure H3: Load Cell Adapter Plate

Counter Sink for the 5/16"-18 flush to the surface
Material Mild Steel

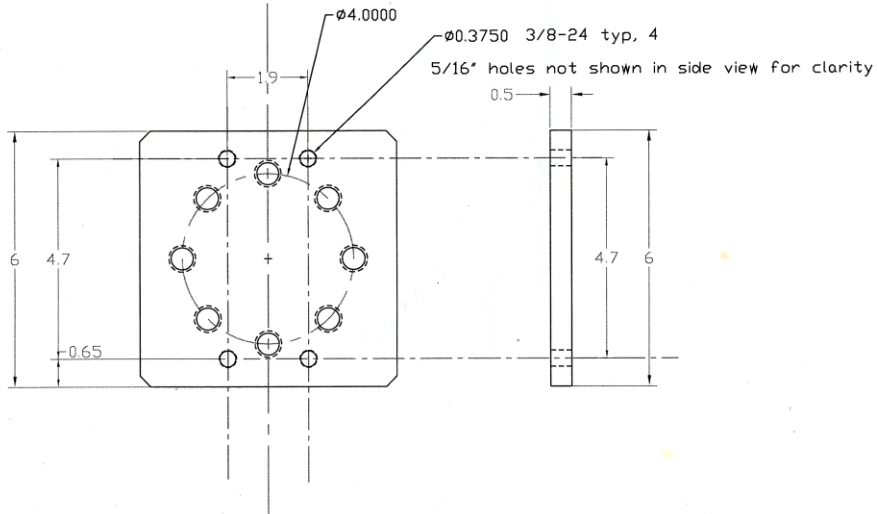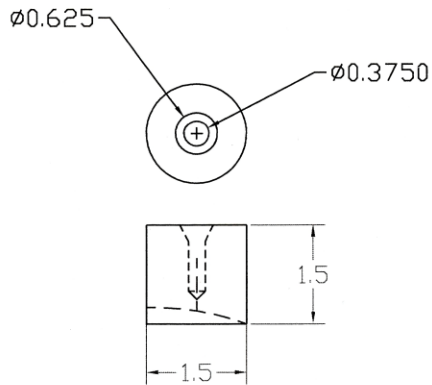5/16"-18, Typ, 8 regularly spaced on a 4.00" DIA circle.

∅4.0000

∅0.3750 3/8-24 typ, 4

5/16" holes not shown in side view for clarity

1.9

0.5

6   4.7

4.7   6

0.65

Figure H4: Bolt Connector

∅0.625

∅0.3750

Bolt is 3/8"-16
Inner Hole Diameter is Clearance for bolt
Inner hole is blind NOT THROUGH DRILLED
Countersink should help feed bolt into recess

1.5

1.5

Aluminum disks epoxied onto helmet – 3 per Helmet
Underside curved to fit helmet

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