FREE VIBRATION RESPONSE OF THE HUMAN HEAD AND NECK SYSTEM UNDER NORMAL AND MOUTHGUARD CONDITIONS

By

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Ethics Approval

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Abstract

Sport-related head injury is considered an important public health problem due to its high number of incidences. Changes in equipment designs have been shown to reduce the incidence of brain injury in sports such as hockey, football, and boxing. One measure that is currently being advocated is to use mouthguards to further help decrease the rate of concussion in sports. There is now a fair body of knowledge indicating their success with respect to reduction of dental injuries. However, no human study appears to have investigated the ability of mouthguards to attenuate force directed upwards across the jaw to the head (an upper-cut blow). Thus the objectives of this study were to develop an experimental setup that would allow for the measurement of the head-and-neck system's free vibration, and to evaluate the *in situ* effects of different mouthguards on this system.

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I would like to dedicate this study to my father Deok Young Lim (1945-2001). Thank you dad. I miss you so much.

Lastly, but not least. I would like to thank God. I now realize that you were the one that carried me through probably the most emotionally and spiritually draining part of my life. I pray and hope that even through this study your name may be glorified. Amen.

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Chapter 1: Review of Literature

1.1 Introduction

With over a third of the Canadian population aged 15 years and older participating regularly in sport, as reported by Canada Statistics, athletes who choose to participate in sports place themselves at risk for injury (Canada Statistics, 1998). The high number of head injuries in sport is being viewed as a cause for concern. The concussed athletes who return after recovery are not only at risk of re-injury but also to the effects of repeated concussive blows. The diagnosis of concussion primarily depends on the symptoms reported by the athletes. However, the wide range of symptoms associated with concussion could be reflective of the numerous parts of the traumatized brain. The potential seriousness is increasingly being recognized through continuing research in the area of sport-related head injuries.

1.2 Incidence Rate

Sport-related head injury is considered an important public health problem due to its high number of incidences (Bailes and Cantu, 2001; Powel and Barber-Foss, 1999; Powel, 2001; Ruchinskas, et al., 1997; Thurman, et al., 1998). In Thurman et al.'s (Thurman, et al., 1998) study, it has been reported approximately 306 000 brain injuries were attributed to sports or other physical activity in the United States during a 12 month period. Sport-related brain injury accounts for approximately 20% of all brain injury. A limitation to such studies is the biased estimates that may have risen from data collection.

Both under-reporting and over-reporting could have occurred. Nevertheless, the large numbers reported indicate that sport-related brain injury is a large public health problem.

To further assess sport-related risk factors, the incidence rates of head injury in various sports have been collected. Some sports that have been identified as potential risk of sustaining head injury are: boxing, football, soccer, rugby, ice hockey, wrestling, and basketball (Bailes and Cantu, 2001; Powel and Barber-Foss, 1999; Powel, 2001; Ruchinskas, et al., 1997; Thurman, et al., 1998). It has been shown that football accounts for 63% of all reported head injury in high school sports in the United States (Powel and Barber-Foss, 1999). In another study conducted in British Columbia, Canada, an incident rate of 5.95 per 1000 player/game hours has been reported among Junior A hockey (Goodman, et al., 2001).

Data compiled from four states by David Thurman et al. (Thurman, et al., 1998) indicates that rates of traumatic brain injuries were highest in 15 to 24 age groups. This is a concern since the quality of life of the injured athletes is potentially jeopardized due to prolonged symptoms of concussion. In addition, rates of brain injury were highest in males, in all age groups, compared to the females (Thurman, et al., 1998). It is not surprising to see higher rates of injuries among males since more males are involved in contact and extreme sports. In female contact sports, the rules are emphasized more toward decreasing the chance of such injuries; whereas with males, the rules are less strict. For example, in women's lacrosse, cross-checking is not allowed. Whereas, in men's lacrosse, cross-checking is used very often and can potentially lead to an increased number of orofacial injuries. Hence, one should be prepared to see more brain injuries in male sports.

1.3 Mouthguards

Due to the growing concern in concussion in sports, numerous studies have investigated this problem and have led to the associated changes in rules and improvement of equipment design (Bailes and Cantu, 2001). Changes in helmet designs have been shown to reduce the incidence of brain injury in sports such as hockey, football, and boxing. One measure that is currently being advocated is to use mouthguards to further help decrease the rate of concussion in sports (Benson, et al., 2002; Biasca, et al., 2002; McCrory, 2001; McCrory, 2001). It is thought that similar properties of mouthguards that provide protection against maxillo-facial injuries may also provide protection against concussion. Although this idea has been presented in numerous papers it has yet to be supported from human studies.

Mouthguards, or mouth protectors, are appliances placed inside the mouth of athletes to reduce mouth injuries. Mouthguards were originally developed in 1890 by Woolf Krause as a means of protecting boxers from lip lacerations (Lawrence and Ward, 1994). Since then mouthguards have been used in most contact sports and their protective function includes more than lip laceration prevention. It has been shown that they offer athletes protection against fractures and dislocations of the jaw, and protection against injuries to the teeth (Chalmers, 1998; Hoffmann, et al., 1999; Newsome, et al., 2001). Sports-related occurrence of dental trauma is reported to be as high as 39% of all patients with dental trauma (Camp, 2000; Gassner, et al., 1999). Numerous studies have shown that mouthguards can reduce the occurrence of dental injuries by means of force absorption and force distribution (Bemelmanns and Pfeiffer, 2001; Bulsara and Matthew, 1998; Chalmers, 1998; Guevara, et al., 2001; Hoffman, et al., 1999; Morikawa, et al., 1998; Newsome, et al., 2001; Warnet and Greasley, 2001). To this day, protection against dental injury is emphasized as the main benefit of using mouthguards.

Since the introduction of mouthguards into contact sports, there have been numerous advances in the manufacturing techniques and materials used for mouthguards. The most commonly used material for mouthguard fabrication, for both boil-and-bite and custom-fitted mouthguards, is the polyvinylacetate polyethylene co-polymer (EVA). EVA is also commonly used in the sole of athletic shoes as a means to absorb shock. Other materials that have been used for mouthguard fabrication are polyvinylchloride, natural rubber, soft acrylic resin and polyurethane (Bulsara and Matthew, 1998; Jagger, et al., 2000; Tran, et al., 2001; Westerman, et al., 1997; Westerman, et al., 2000).

Mouthguards have been classified into 3 general types by the American Society for Testing and Materials (American Society for Testing and Materials):

stock (Type I)

mouth-formed (Type II)

custom-fitted (Type III)

1.3.1 Stock Mouthguards (Type I)

Type I mouthguard was the first type to become commercially available and used in sports. Type I mouthguards are purchased over-the-counter and are 'ready-made;' they do not require further modification. However, they are considered to be the least satisfactory form of mouthguard because they are bulky and ill-fitting. They are held in place by clenching the teeth together, and interfere with talking and breathing (Chalmers, 1998; Maestrello, et al., 1999). Consequently, the use of Type I mouthguards is actively being discouraged (Chalmers, 1998; Maestrello, et al., 1999). Although they do cost the least to purchase, the prices of some type II mouthguards are low enough (under $5 CDN) that the cost is no longer considered an issue. In addition, most sporting-goods store no longer sell Type I mouthguards.

1.3.2 Mouth-Formed Mouthguards (Type II)

Type II mouthguards, more commonly known as 'boil-and-bite,' are relatively inexpensive and can also be purchased over-the-counter. Type II mouthguards are made from a preformed thermoplastic polyvinylacetate-polyethylene co-polymer (Barth, et al., 1996; Hoffman, et al., 1999; Newsome, et al., 2001). It is softened by immersion in boiling water for 10 seconds; it is placed in the mouth and moulded to the teeth of the wearer by using finger, tongue and biting pressure. One major concern with Type II mouthguard is the occlusal thinning that occurs during the moulding process (Hoffman, et al., 1999, Tran, et al., 2001). There is no control over the final thickness of the mouthguard, which compromises the protectiveness of the mouthguard. The cost of Type II mouthguard ranges from $5 CDN to $30 CDN.

1.3.3 Custom-Fitted Mouthguards (Type III)

Type III mouthguards are custom-made over a cast model of the wearer. They are the most expensive type as they are custom-made by dentists. The dentist takes an impression of the wearer's maxillary arch (the upper teeth) and uses it to build a cast model, on which a thermoplastic material is formed. Unlike the Type II mouthguards, the thickness of custom-made mouthguards can be monitored and controlled throughout the fabrication process. Type III mouthguards are fabricated from a thermoplastic co-polymer, most commonly polyvinylacetate polyethylene (Hoffman, et al., 1999; Newsome, et al., 2001). The cost of Type III mouthguard ranges from $150 CDN to $450 CDN.

1.3.4 Maxillary and Bi-maxillary Mouthguards

Mouthguards are usually fitted and worn on the maxillary arch only. The athletes also have the option of purchasing bi-maxillary mouthguards for both Type II and Type III mouthguards. Bi-maxillary mouthguards are constructed in one piece and covers both the maxillary and mandibular arches. Some of the benefits of bi-maxillary mouthguards are: complete protection for all of the teeth, reduced interference with airflow, and increased energy absorption and dissipation (Milward and Jagger, 1997; Newsome, et al., 2001). However, bi-maxillary mouthguards costs nearly twice as much as the maxillary mouthguards.

1.4 Recent Studies

It is generally accepted that the mouthguard's main function is primarily to provide injury prevention to teeth. Therefore, most mouthguard studies examine this protective nature of different mouthguards and mouthguard materials (Bemelmanns and Pfeiffer, 2001; Bulsara and Matthew, 1998; Greasley and Karet, 1997; Guevara, et al., 2001; Hoffmann, et al., 1999; Jagger, et al., 2000; Tran, et al., 2001; Warnet and Greasley, 2001; Westerman, et al., 1997; Westerman, et al., 2000). The studies that test different types of mouthguards use a simulated maxillary arch, made from a rubber arch containing replaceable ceramic teeth and a renewable composite jawbone. The simulated maxillary arch is fitted with a mouthguard and an impact device is used to deliver a horizontal blow to the maxillary arch. Tooth displacement and tooth fractures caused by the projectiles are often used as indices for assessing different mouthguards. Although this method provides useful information when comparing mouthguards, a drawback is that the simulated maxillary arch is not a biofidelic model of the human. Ideally, the maxillary arch should be built to accurately simulate the soft tissue properties of the oral cavity and the surrounding tissues. Hence, the results may not reflect the protective value of the mouthguard under realistic loading conditions.

From the studies that investigated various mouthguard materials and designs, it was found that the individual cushioning effect, energy absorption, are directly correlated to the thickness of the material used (Figure 1) (Bemelmanns and Pfeiffer, 2001; Hoffmann, et al., 1999; Tran, et al., 2001). When fitting the boil-and-bite mouthguards, there is always a thinning of the material in the occlusal surfaces. It is hypothesized that the protective nature of mouthguards against concussion lies in the ability of the mouthguard to absorb energy when hit across the occlusal surface. Therefore, the fitting process for the boil-and-bite mouthguards is of some concern. In this light, the use of properly fabricated custom-fitted mouthguard is encouraged, since the thickness can be controlled.

It is a generally accepted that custom-fitted mouthguards are better than boil-and-bites (Chalmers, 1998; Guevara, et al., 2001; Maestrello, et al., 1999; Newsome, et al., 2001; Tran, et al., 2001). Athletes playing in more vulnerable positions should be encouraged to wear custom-fitted mouthguards. However, some of these recommendations are not based on experimental evidence, but on characteristics that differentiates the two types of mouthguards instead. For instance, custom-fitted mouthguards are frabricated to enclose all teeth, whereas, boil-and-bite mouthguards cannot be customized to individual players. Therefore, it is believed that custom-fitted mouthguards provide more protection. Despite popular beliefs, recommendation of mouthguards should be based on scientific evidence.

In Bemelmanns and Pfeiffer's study (Bemelmanns and Pfeiffer, 2001), they investigated the shock absorption capacities of boil-and-bite and custom-fitted mouthguards. For boil-and-bite mouthguards, the force attenuation capacities were 18.9%, 16.9%, and 13.3% at 250N, 350N, and 500N respectively. For custom-fitted mouthguards, the absorption capacities were 33.3%, 33.2%, and 25.7%. The study provides good evidence for the superior protection offered by the custom-fitted mouthguards. However, a shortcoming to this study is that only one brand of boil-and-bite mouthguard was tested against six different types of custom-fitted mouthguards. The data from the boil-and-bite mouthguard may not be a realistic representation of the boil-and-bite mouthguard population.

Guevara et al. (Guevara, et al., 2001) compared the number of fractured ceramic teeth resulting from a striking pendulum, when fitted with a boil-and-bite or custom-fitted mouthguard. Contrary to their expectations, all ten ceramic teeth fractured when the custom-fitted mouthguards were tested, compared to no fracture for two boil-and-bite brands and two fractured teeth for the third boil-and-bite brand that were tested. These unexpected results could be attributed to the observed difference in thickness. The boil-and-bite mouthguards were over twice as thick than the custom-fitted mouthguards. Guevara and his colleagues used a standard 3.0mm polyvinylacetate polyethylene sheet and noticed that it was reduced to 2.0mm after the mouthguard was formed. Although custom-fitted mouthguards are more comfortable to wear, unless the mouthguard is fabricated properly and the thickness is monitored, custom-fitted mouthguards might be less protective than boil-and-bites. Dentists and laboratory technicians must be made aware of the importance of controlling the final thickness of mouthguards in order to provide superior protection.

1.5 Biomechanics of Impact and Injury

Impact response of two colliding bodies depends on the mechanical properties of both bodies. In a collision, an object with a given mass experiences a force for a specific amount of time which results in a change in momentum. The impulse experienced by the object equals the change in momentum of the object (Force*time = mass*change in velocity). Decreasing the modulus and mass of the impacting object has been shown to decrease the peak force sustained by the body being impacted on (Crisco, et al., 1996). In the same way, three factors influencing the impact force on the upper and lower extremities of humans are surface stiffness and damping properties, and the effective mass (DeGoede, et al., 2002; Robinovitch, et al., 1991; Robinovitch, et al., 1997). Since the degree of injury is related to the magnitude of the impact force, lowering this impact force should be a priority in injury prevention. One way to assess the protection offered by sporting equipment is to examine their effect on system stiffness and damping, which in turn affects peak force.

To objectively test the potential protective value of mouthguards against concussion, the biomechanics of brain injury must be considered. For the first half of the twentieth century, most authors believed that focal loading of the brain (coup phenomena) was the major injuring factor involved in brain damage (Figure 2) (Gennarelli, 1982; Sahuquillo, et al., 2001). Recently, the concept that the generation of shear strain throughout the brain is an important cause of brain dysfunction has been studied and published (Christopher, 1993; Lawrence and Ward, 1994; McIntosh, et al., 1996; Nishimoto and Murakami, 1998; Whiting and Zernicke, 1998). Injury to the brain can be caused by forces applied to the head and also by the resulting abrupt motions transmitted to the head. Hence, the focal loading theory can be regarded as being "incomplete" because it does not address the injury arising without direct contact to the head. Consequently, the proposed study will focus mainly on the centripetal theory of brain injury (Lawrence and Ward, 1994). The centripetal theory states that rotational acceleration can produce both focal and diffuse brain injuries; hence, it is more deleterious than translational acceleration.

When the head suffers a sudden mechanically induced load, two different types of acceleration can be induced: translation and rotation (Cantu, 1996; Gennarelli, 1982; Lawrence and Ward, 1994; Whiting and Zernicke, 1998; Zhang, et al., 2001). Translation means that the head's center of gravity moves along a straight line and rotation means that the head moves around its centre of gravity. There are three distinct types of stress that can be generated by these two acceleration forces to the head. The first is compressive, the second is tensile (the opposite of compressive)—the first two stresses are associated with translational acceleration. The third is shearing (a force applied in parallel to a surface), which is mostly associated with rotational acceleration (Lawrence and Ward, 1994; Nishimoto and Murakami, 1998; Runnerstam, et al., 2001). If the magnitude of stresses induced in the tissues is sufficiently great, the tissues will fail and injury will occur. However, shearing forces are poorly tolerated in brain tissue, whereas compressive and tensile forces are relatively well tolerated (Lawrence and Ward, 1994; Nishimoto and Murakami, 1998; Runnerstam, et al., 2001). Consequently, many authors consider rotational forces to be more damaging than translational forces. Gennareli et al. (Gennarelli, 1982) demonstrated that translation of the head in the horizontal plane produced essentially only focal effects, while diffuse injuries were seen only when a rotational component was present. In his studies, monkeys were either subjected to pure translational acceleration or pure rotational acceleration. Even at maximal acceleration, translational acceleration did not induce cerebral concussion; whereas, it was easy to induce concussion from rotational acceleration.

There are two physiological mechanisms by which the brain is protected: the skull and the cerebrospinal fluid (CSF). The skull is the bony tissue that encases the brain and provides structural integrity. The CSF is a fluid that surrounds the brain and acts as a shock absorber. The focally applied stress is absorbed and distributed in a uniform fashion. However, the CSF does not totally prevent shearing forces from being transmitted to the brain, especially in the case of rotational forces (Cantu, 1996). In addition, shearing forces are maximal where rotational gliding is hindered within the brain (Zhang, et al., 2001). These regions are where tissues with different stiffness come into contact—for example, the brainstem-cerebrum attachments.

Diffuse axonal injury (DAI) is characterized by widespread injury to the cerebral white matter, the brainstem, and the corpus callosum (Arbogast and Susan, 1998; Arbogast and Susan, 1999; Besenski, 2002). DAI is caused by stretching and tearing of nerve supplies and small blood vessels. Numerous authors have investigated, mathematically and experimentally, the effects of rotational acceleration on these structures (Arbogast and Susan, 1998; Besenski, 2002; Cantu, 1996; Nishimoto and Murakami, 1998; Runnerstam, et al., 2001). In these studies, the regional strains showed that the brainstem and corpus callosum regions experienced significantly higher strains during rotational loading. Although brain injury is a combination of coup, contra-coup and rotational forces, there is sufficient evidence to indicate the negative outcome of pure rotational acceleration on the brain. Jaw impact could produce concussion because it produces rotational (shear) forces on the brain. For this reason, the proposed study will investigate the protection that mouthguards could potentially provide against injury to the brain.

1.6 Mouthguards and Rotational Acceleration

Recently, numerous investigators have suggested that mouthguards could potentially reduce the occurrence of concussion in athletes (Benson, et al., 2002; Biasca, et al., 2002; McCrory, 2001; McCrory, 2001). However, there is yet no convincing evidence to support this claim. Various theories, addressing how a mouthguard might provide protection against concussion, have been developed. Through radiography, it has been shown that mouthguards alter mandibular position, so that the condyles are distracted away from their fossae (Biasca, et al., 2002; McCrory, 2001). As a result, forces from mandibular impact that would normally be transmitted directly from the condylar heads to the cranium will be attenuated by the increased gap. In addition, it is proposed that a mouthguard could act by absorbing some of the impact force received from a blow on the inferior aspect of the mandible. Therefore, the rotation effect of the impact would be lessened. If the rotational acceleration can indeed be reduced, it would make a good case that mouthguards can reduce the occurrence of concussions.

1.7 Rationale, Purpose and Hypotheses

1.7.1 Rationale

Mouthguards are primarily used for protection of dentition against trauma. Therefore, they focus on impacts that could potentially cause injury to the dentition (a direct hit to the teeth). As seen in numerous reports, with the improvement in mouthguard manufacturing techniques as well as the upgrading of the materials used, the ability of mouthguards to attenuate shock has increased and the rate of dental injuries has indeed decreased. However, no human study appears to have investigated the ability of mouthguards to attenuate force directed upwards across the jaw to the head. An upward force placed inferior to the mandible would result in the rotation of the head about the neck. Although both translational (coup and contra-coup) and rotational acceleration in various planes causes injury to the brain, the results from the proposed study will be applicable to the rotational inertia in the sagittal plane. It is proposed in numerous papers that mouthguards should theoretically provide more protective against rotational acceleration than translational acceleration.

Since the mouthguard lies within the occlusal surface, between the upper and lower set of teeth, one could hypothesize that mouthguards might attenuate force that is delivered across the jaw to the head. The proposed study will explore this potential aspect of mouthguards. This could be clinically important since a blow to the chin, like any other force directed towards the brain, can result in a concussion. Hence, the properties of mouthguards (the manufacturing techniques and the materials used) that decrease the force transmitted to the skull may also provide some protection from brain injury.

It is important to develop a mechanical testing system for evaluating the protective value of mouthguards under realistic loading conditions. However, no relevant mathematical model exists which can be used to construct a bio-fidelic mechanical testing system. Testing on human subjects must be done in order to determine the design parameters for a testing system. Until further testing can be done on a computer-based model with realistic loading conditions, the collected data from this proposed study could be used to extrapolate mouthguard performance under high-energy conditions.

This study addresses questions related to the force attenuation characteristics of mouthguards. This may be applicable to when an athlete is hit with an upper-cut blow to the jaw. Our results should provide insight on the potential benefits mouthguards provide in reducing the probability of brain injury in athletes.

1.7.2 Purpose

To assess how the impact response (stiffness and damping values) of the jaw and neck are affected by the use of mouthguards.

To compare different boil-and-bite mouthguards, with varying material and manufacturing properties, in controlled conditions involving specific upper limb orientations, and excitation weight.

To develop an experiment that will measure the effective mass of the head.

To compare the head mass values obtained from the experiment to the values calculated from an anthropometric measurement method.

1.7.3 Hypotheses

With the introduction of a mouthguard into the jaw and head system, system stiffness (k) will decrease.

Mouthguards will differ in their ability to decrease system stiffness (k). This is due to different designs and the different materials used in the fabrication of mouthguards.

System stiffness (k) will be higher in "arms un-supported" condition when compared to "arms supported" condition. While speculative, it is thought that "arms un-supported" action incorporates more musculature; thus, increasing the system mass.

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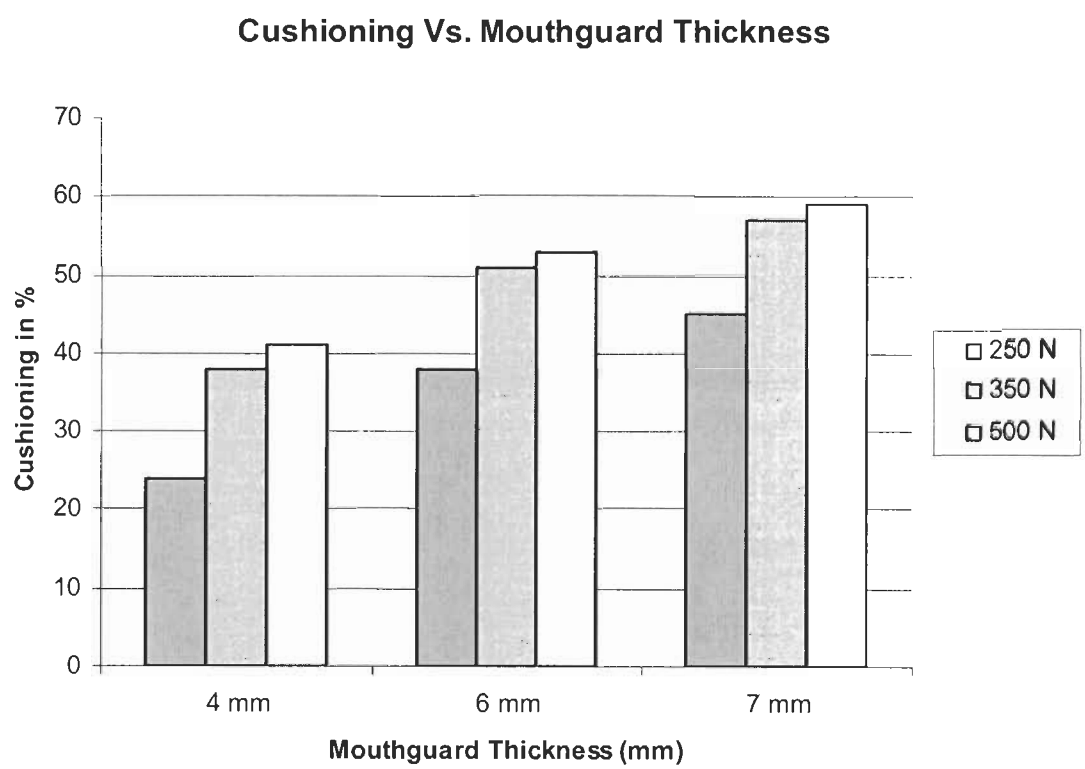
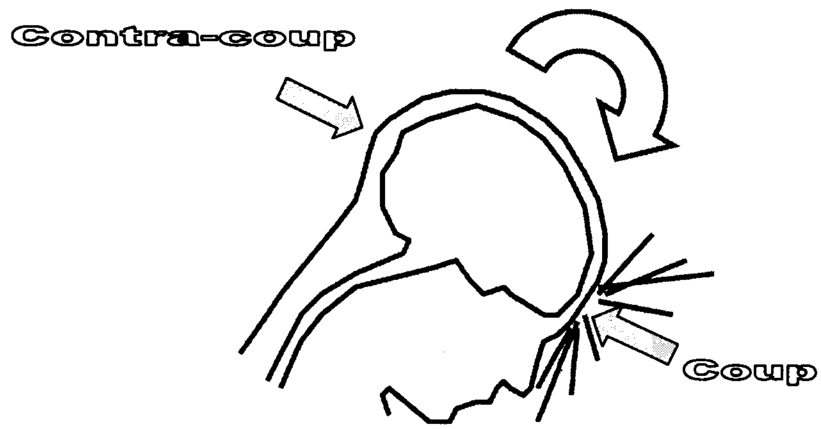
Figure 1: Effects of thickness on force attenuation



Data source: Hoffmann, et al., 1999

Figure 2: Coup and Contra-coup Head Injury



Chapter 2: Technical Note - Effective Mass of the Human Head

2.1 Abstract

Conclusion: In experiments that require the mass of the head to simulate impact situations, the effective mass of the head should be determined. A new method of obtaining the effective mass is through the free vibration technique used in this study. **Purpose:** The purpose of this study was to develop a technique for measuring of the effective mass of the human head during impact to the jaw. **Method:** To calculate the head's effective mass, the undamped natural frequency of the head was measured under normal and free vibration conditions. The head's free vibration response for each condition was established with a change in vertical force applied at the chin. The participants' (n=12) heads were perturbed by the sudden and unexpected removal of an excitation weight which rested on their heads. Three different predetermined excitation weights were used: 11.5N (1.2 kg), 16.5N (1.7 kg), and 20.8N (2.1 kg). Data were collected with a load cell placed in series with a chin platform. To determine the effect of upper limb support conditions, two arm positions were used: arm-supported and arm-unsupported. The mass of the head estimated by Clauser's 1969 regression was compared with predictions from the technique. **Result:** No difference in the undamped natural frequency was observed between arms-supported and arms-unsupported conditions, or between the three different excitation weights. The effective mass, as determined by the

free vibration experiment, was significantly higher than the estimated value from Clauser's 1969 regression ($p < .001$).

2.2 Introduction

Biological tissues during impact can be considered as a mechanical system with masses connected by springs and dampers. These model parameters must be properly identified in order to accurately simulate an impact to this system. Two diverse methods have previously been employed to obtain the segment specific masses. The theoretical approach uses calculations based on anthropometric measurements, while the empirical approach uses direct measures. Once a technique to estimate the model parameters of the head is developed, it can be used to test helmet and mouthguard designs, and to predict injury risk.

In 1969, Clauser, McConville, and Young developed a series of equations used to predict masses of specific segments of the human body (Marfell-Jones, 1984). Like many other similar studies, the authors based their research on cadavers (n=12). The best predictor of segmental masses of the human body is total body weight (Marfell-Jones, 1984). Not surprisingly, with the inclusion of segment specific anthropometric measurements in the formula the prediction error is reduced. Clauser's equations are still being used today in biomechanical studies (Nigg and Liu, 1999), where for the prediction of the mass of the head Clauser's regression equation uses two variables: total body weight, and head circumference.

Another method used in biomechanical studies to estimate segment masses is perturbation experiments (Lin, et al., 2001; Robinovitch, et al., 1997). However, there has yet to be a study that has developed a technique to apply this method to estimate the mass of the head. A benefit of this method over the anthropometric method is that it measures the effective mass whereas the anthropometric method estimates the 'static' mass. The effective mass is the total moving mass with respect to a specific location during a movement, and is affected by other intrinsic properties such as muscle tone. Postural neck muscles stabilize the head unto the torso, essentially increasing effective mass of the system. When studying impact situations, it is more appropriate to use the effective mass because it simulates realistic conditions (Liu and Nigg, 2000).

The purpose of this study was to design an experiment to measure the effective mass of the head. The second goal was to compare the free vibration technique to the values estimated by Clauser's formula. The undamped natural frequency of the head was measured under normal and free vibration conditions. The perturbation of the head was initiated by the removal of an excitation weight initially placed on the participant's head. The effect of arm support and different excitation weights on the calculated mass of the head was also measured. The effect of different excitation weights were looked at to test system linearity (whether effective mass changed with the amplitude of vibration). In addition, the estimated mass of the head from the anthropometric method was compared with the free vibration method. This study also looked at whether the measured effective mass of the head associates with total body mass and head circumference, as suggested by Clauser. It is hoped that future studies may use this technique to develop better helmet and mouthguard designs, and to predict injury risk. In a subsequent study this technique was used to compare various mouthguards.

2.3 2.3 Methods

2.3.1 Subjects

12 male subjects between the ages of 19 and 28 participated in this study (Table 1). Those who showed interest in participating in the study were asked to fill out a Physical Activity Readiness Questionnaire (PAR-Q) and an informed consent was obtained. Ancillary measures on the subjects' weight and head circumference (forehead circumference) were taken. These measured values were used to estimate the mass of the head using Clauser's 1969 regression formula.

2.3.2 Human Head Model

Head perturbation experiments measure the dynamic response of the head and neck to an applied step change in vertical force applied to the chin. The simplest model capable of simulating the head during the experiment consists of a single effective mass attached to a parallel arrangement of spring and damper (k and b) (Figure 3). The effective mass represents the moving mass of the head and the spring-damper element is mainly associated with the muscles and ligaments surrounding the jaw and neck.

2.3.3 Experimental Protocol

To calculate the effective mass of the head (m) we needed to conduct two experiments with each participant. Both experiments involved measures of the head system's free vibration response to a step change in applied vertical force, but were distinguished by different support conditions at the chin.

Support condition 1 was the "conventional" head perturbation experiment used in our mouthguard testing experiment. Subjects sat erect on a stool and placed their chins on a chin platform. Plasticine was glued on to the chin platform and formed to fit the participants' chins. The chin platform attached to a load cell (Sensotec® Tension/Compression load cell Model 41), which was used to record the changes in force across the load cell. The load cell was secured on a tabletop. To allow each subject to sit uniformly, the height of the table was adjusted with the use of a hydraulic pump. An excitation weight was placed on the apex of the participants' head, and once the signal tracing had stabilized for five seconds, the excitation weight was quickly removed and data were collected for five seconds (DATAQ™) at a sample rate of 488 Hz. The excitation weight was lifted by releasing a counter-weight attached to it via a pulley system secured to the ceiling (illustrated in Figure 4). These predetermined excitation weights were: 11.5N (1.2 kg), 16.5N (1.7 kg), and 20.8N (2.1 kg). Prior to the removal of the excitation weight the subjects were instructed to relax their neck muscles as much as possible.

Support condition 2, or the "spring platform" condition, was identical to the conventional experiment, except for the chin was resting on a linear spring platform of stiffness k_s. In the spring platform condition, k_s is placed in series with the parallel combination of k and b (Figure 3). The addition of the spring lowers the measured frequency of vibration, and provides an additional equation to solve for the unknown parameter m (see *Parameter Identification*).

To measure the stiffness of the spring platform (k_s), free vibration tests were conducted on the spring platform. Similarly to the conventional experiment, a static mass of 1.2 kg was placed on the spring platform and an excitation weight was quickly removed from the system. The testing was performed with excitation masses of 1.7 kg and 2.1 kg. The spring platform stiffness k_s was equal to

(1) $$k_s = \omega_n * 1.2 .$$

The calculated stiffness of the spring platform was 4181 N/m.

It was noted in pilot trials that arm support conditions influenced the measured effective mass. The measured effective mass could be influenced by torso or even pelvis movement. In this study two arm positions were used: arms-supported and arms-unsupported. In the arms-supported position, the subjects were instructed to place their arms on the table and use them to support their upper body weight. In the arms-unsupported position the subjects placed their hands on their lap. In both testing conditions the participants sat erect on a chair and rested their chins on the chin platform.

2.3.4 Parameter Identification

In both experiments, the damped natural frequency ω_d and damping ratio ξ were measured. The damped natural frequency was equal to

(2) $$\omega_d = \frac{2\pi}{T_d}$$

where T_d was defined by the damped natural period equal to twice the time interval between the first force maximum and the first force minimum. The damping ratio was approximated by

(3) $$\xi = \frac{\delta}{\sqrt{\pi^2 + \delta^2}}$$

where δ was defined by the logarithmic decrement,

$$(4) \qquad \delta = \ln\left(\frac{F_1}{F_2}\right),$$

where F_1 was the difference between the first force maximum and the end load, and F_2 was the difference between the end load and the first force minimum (Figure 5). The undamped natural frequency was ω_n given by

$$(5) \qquad \omega_n = \frac{\omega_d}{\sqrt{1-\zeta^2}}.$$

From these parameters, the undamped natural frequency ω_n was calculated for both support conditions. In the conventional condition k and m contributed to ω_{n1}, whereas in the spring platform condition k, m, and k_s contributed to ω_{n2}. Therefore, ω_{n1} and ω_{n2} were equal to

$$(6) \qquad \omega_{n1} = \sqrt{\frac{k}{m}}$$

and

$$(7) \qquad \omega_{n2} = \sqrt{\frac{\left(\frac{k * k_s}{k + k_s}\right)}{m}}.$$

By combining Eq. 1, 6, and 7, the effective mass m was given by

$$(8) \qquad m = k_s\left(\frac{1}{\omega_{n2}^{\;2}} - \frac{1}{\omega_{n1}^{\;2}}\right).$$

As a comparison to calculated values of m, we used Clauser's 1969 regression equation to estimate the mass of the head as:

$$m = (0.104 * \text{head circumference}) + (0.015 * \text{total body mass}) - 2.189$$

2.3.5 Data Analysis

The mean undamped natural frequency (ω_n), over 3 repeated trials, for each load x support condition was computed, and a single repeated measures ANOVA on the three excitation weights for each position and platform type was performed. After showing there was no effect of excitation weight the ω_n of the three excitation weights were collapsed together, and the mean ω_{n1} and ω_{n2} values for arm-supported and arm-unsupported were calculated for each participant. From these values the effective mass of the head was computed for both arm positions and compared using a paired t-test. In the third comparison, the effective mass of the head of both arm positions were collapsed together because they had no effect on the calculated variables. The combined value of the effective mass of the head was then compared with the calculated value from the Clauser's 1969 regression using a paired t-test. For all comparisons, the significance value was set at $p < .05$ and all data are presented as mean $\pm$ SD.

To investigate the association of measured effective mass with total body mass and head circumference a correlation matrix and a multiple linear regression was performed.

2.4 Results

2.4.1 Excitation Weight

No significant difference was found between the excitation weights for the conventional platform ($F_{2,33}$=1.78; p = 0.19) (Figure 6). For the conventional platform, in the arm-supported position the undamped natural frequencies were 54.8 ± 8.1 rad/sec, 48.7 ± 9.3 rad/sec, and 49.5 ± 9.5 rad/sec for 1.2 kg, 1.7 kg, and 2.1 kg, respectively. In the arm-unsupported position the values were 54.0 ± 7.9 rad/sec, 51.3 ± 6.6 rad/sec, and 51.8 ± 9.9 for 1.2 kg, 1.7 kg, and 2.1 kg, respectively. No significant difference was found between the excitation weights for the spring platform condition ($F_{2,33}$=0.57; p = 0.58) (Figure 6). For the spring platform, in the arm-supported position the undamped natural frequencies were 22.3 ± 1.8 rad/sec, 22.4 ± 2.0 rad/sec, and 23.8 ± 1.6 rad/sec for 1.2 kg, 1.7 kg, 2.1 kg, respectively. In the arm-supported position the values were 24.4 ± 1.7 rad/sec, 23.6 ± 1.8 rad/sec, and 23.6 ± 1.9 rad/sec for 1.2 kg, 1.7 kg, and 2.1 kg, respectively.

2.4.2 Arm Position

There was no difference in the effective mass of the head between arm-supported (6.02 ± 0.93 kg) and arm-unsupported (5.99 ± 0.70 kg) (mean difference = 0.024; 95% CI = -0.40 to 0.45; p = 0.90).

2.4.3 Mass of Head

A significant difference was found when the mass of the head computed from Clauser's 1969 regression (5.00 ± 0.06 kg) was compared with the spring-platform experiment (6.00 ± 0.24 kg) (mean difference = 1.01; 95% CI = 0.53 to 1.50; p < .001)

(Figure 8). The calculated correlation values for the effective mass against head circumference and total body mass were 0.73 and 0.13, respectively. The multiple linear regression formula for the effective mass was:

effective mass = (0.64*head circumference) – (0.021*total body mass) – 29.1.

2.5 Discussion

As mentioned above, the calculated value of the head's mass from the spring-platform experiment was higher than when using Clauser's 1969 regression equation. Clauser's method estimates the resting mass of the head, whereas the spring-platform estimates the "effective" mass of the head – the mass that reacts to a perturbation. The increase in mass seen with the spring-platform experiment might be due to the activation of the postural neck muscles. Neck muscle tone stabilizes the head unto the torso, essentially increasing the mass of the system. Other factors such as the movements of the neck and torso could also contribute to the value of m. In addition, inter-subject variability is expected in the effective mass of the head. It is recommended that any studies requiring the effective mass of the head obtain it experimentally for each participant.

The free vibration technique developed in this study could be used to test various protective appliances, such as mouthguards. Current mouthguard studies use a simulated maxillary arch, made from a rubber arch containing replaceable ceramic teeth and a renewable composite jawbone (Bemelmanns and Pfeiffer, 2001; Bulsara and Matthew, 1998; Greasley and Karet, 1997; Guevara, et al., 2001; Hoffmann, et al., 1999; Jagger, et al., 2000; Tran, et al., 2001; Warnet and Greasley, 2001; Westerman, et al., 1997;

Westerman, et al., 2000). The simulated maxillary arch is fitted with a mouthguard and an impact device is used to deliver a horizontal blow to the upper jaw. Tooth displacement and tooth fractures caused by the projectiles are often used as indices for assessing different mouthguards. Although this method provides useful information when comparing mouthguards, a drawback is that the simulated maxillary arch is not a biofidelic model. Hence, the results do not necessarily simulate conditions that are seen in clinical observations. The free vibration technique is a novel method by which investigators can obtain spring and damping values of the head-neck system and develop a biofidelic model to test mouthguards.

The excitation weight did not have an effect on the calculated undamped natural frequency for arm-supported and arm-unsupported conditions for both conventional and spring platforms. Therefore, any future studies trying to determine the effective mass of the head with the spring-platform experiment can use a single excitation weight between 1.2 kg and 2.1 kg.

Since no difference was seen in the effective mass of the head between arm-supported and arm-unsupported, either arm position can be used in future studies. This indicates that the source of the moving mass is located above the shoulders.

Graphs of effective head mass vs. head circumference and effective head mass vs. total body mass appear to support the notion that effective mass correlates with head circumference, but not with total body mass (Figure 9 and 10). This suggests that segment masses are better predicted with segment specific measurements such as head circumference, as compared to a general anthropometric measurement such as the total body mass.

2.6 Conclusion

In this study, the effective mass of the head of 12 participants was determined using a free vibration technique. Different excitation weights between 1 and 2 kg did not affect the undamped natural frequency of the system. In addition, the two arm positions that were used did not affect the computed effective mass of the head. In determining the effective mass of the head using the spring-platform experiment, one excitation weight between 1.2 and 2.1 kg and one arm position is adequate. The spring-platform experiment is unique in that it estimates the total moving mass of the head in response to a perturbation. Previous methods rely on ancillary measures to estimate the static mass of the head, therefore should not be used in experiments where dynamic responses of the head are involved. Results from this study suggests that head circumference is a better predictor of effective head mass than total body mass.

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Westerman, B., Stringfellow, P.M., and Eccleston, J.A. The effect on energy absorption of hard inserts in laminated EVA mouthguards. *Australian Dental Journal.* 45(1): 21 -23, 2000.

Table 1: Parameter values and predicted head weight for each subject.

Subject	Head mass (kg) Spring-platform technique	Head mass (kg) Clauser's 1969 regression	ω_{n1^*} (rad/sec) [conventional platform]	ω_{n2} (rad/sec) [spring platform]	Head circumference (cm)	Body mass (kg)
1	5.23	5.09	46.8	24.8	57.0	90
2	4.60	4.68	49.4	26.1	56.5	66
3	5.79	5.07	53.8	24.9	57.5	85
4	7.19	5.07	44.0	21.2	59.0	75
5	5.77	4.67	49.2	23.6	57.0	62
6	6.70	5.04	49.2	22.3	58.0	80
7	7.44	5.31	54.6	21.8	60.0	84
8	6.18	5.00	53.5	23.4	57.0	84
9	4.96	5.13	51.2	25.3	58.0	86
10	6.15	5.04	61.7	24.0	58.0	80
11	5.92	4.63	53.3	23.8	56.5	63
12	6.15	5.21	54.3	23.5	59.0	84
	6.01±0.84	5.00±0.22	51.75±4.54	23.73±1.45	57.79±1.10	78.25±9.56

* Averaged over the three excitation weights

Figure 3: Schematic of the under-damped spring-dashpot system with one degree of freedom.

The model on the upper left shows individual k and b of the jaw (j) and neck (n). This model can be reduced to a single spring-damper-mass system (shown on the upper right). For the spring-platform technique this simplified model is attached in series to a single spring (shown on the lower right).

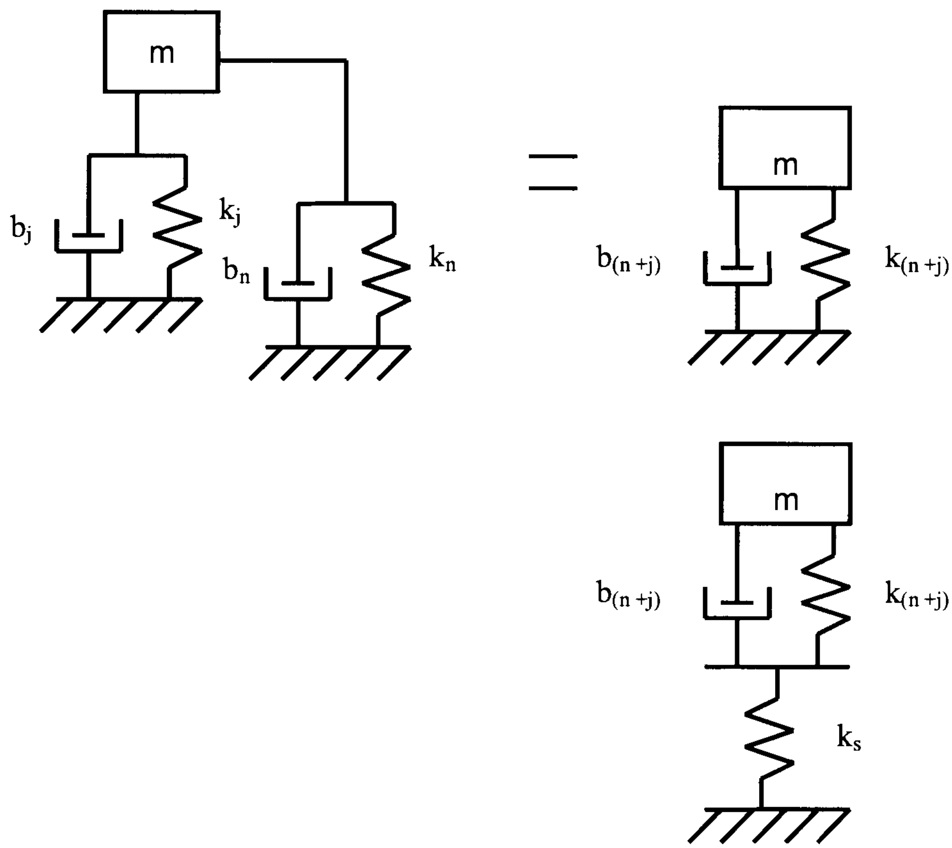


Figure 4: Setup of experiment. When the counter-weight is released the excitation weight is lifted off the head.

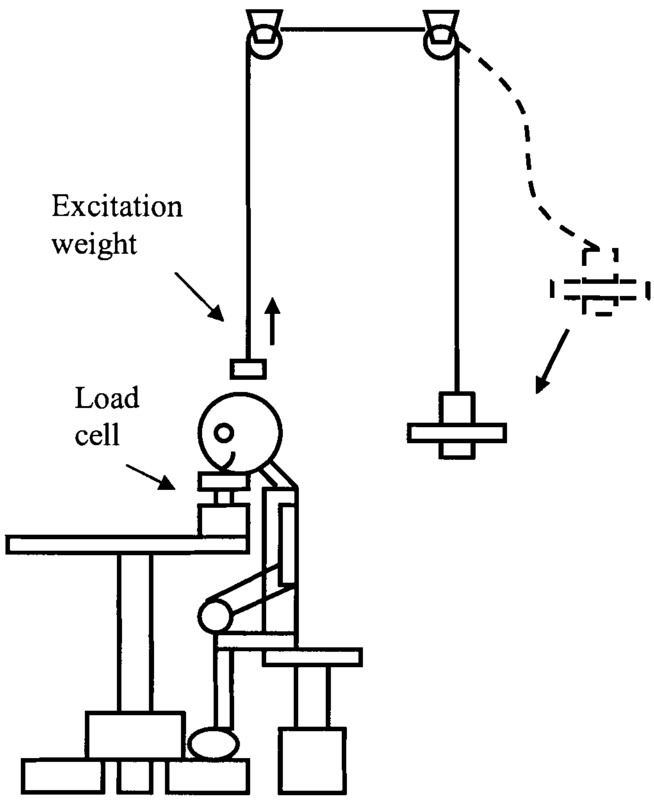


Figure 5: One trial output from DATAQ.

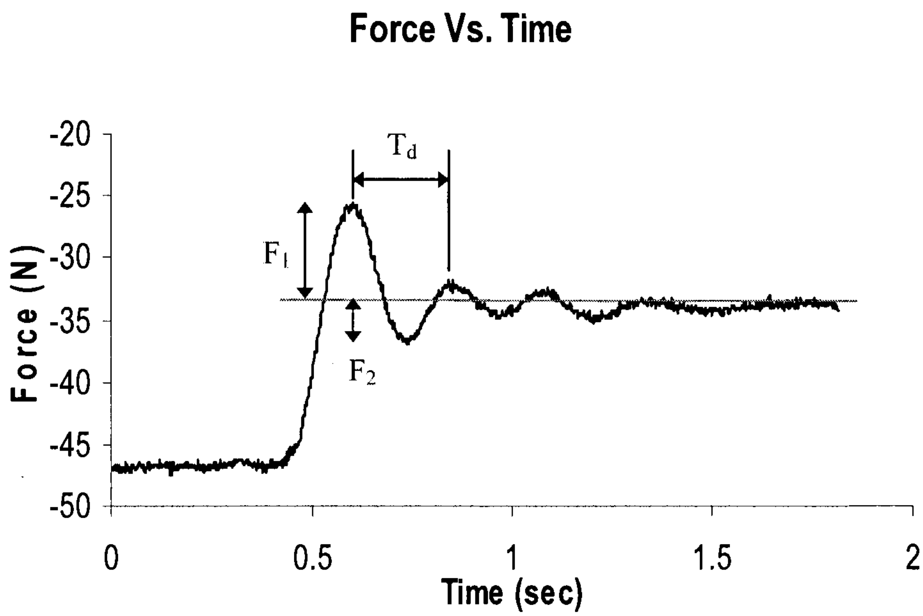


Figure 6: Composite score of calculated undamped natural frequency of different excitation weights for the conventional platform, for arm-supported and arm-unsupported conditions (n=12).

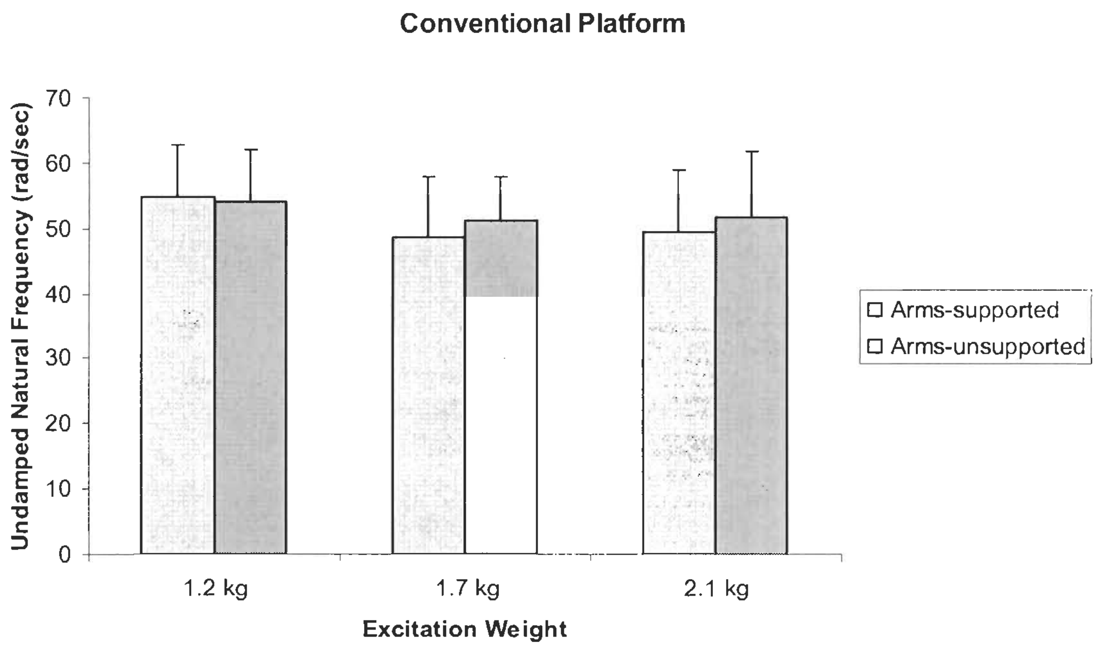


Figure 7: Composite score of calculated undamped natural frequency of different excitation weights for the spring platform, for arm-supported and arm-unsupported conditions (n=12).

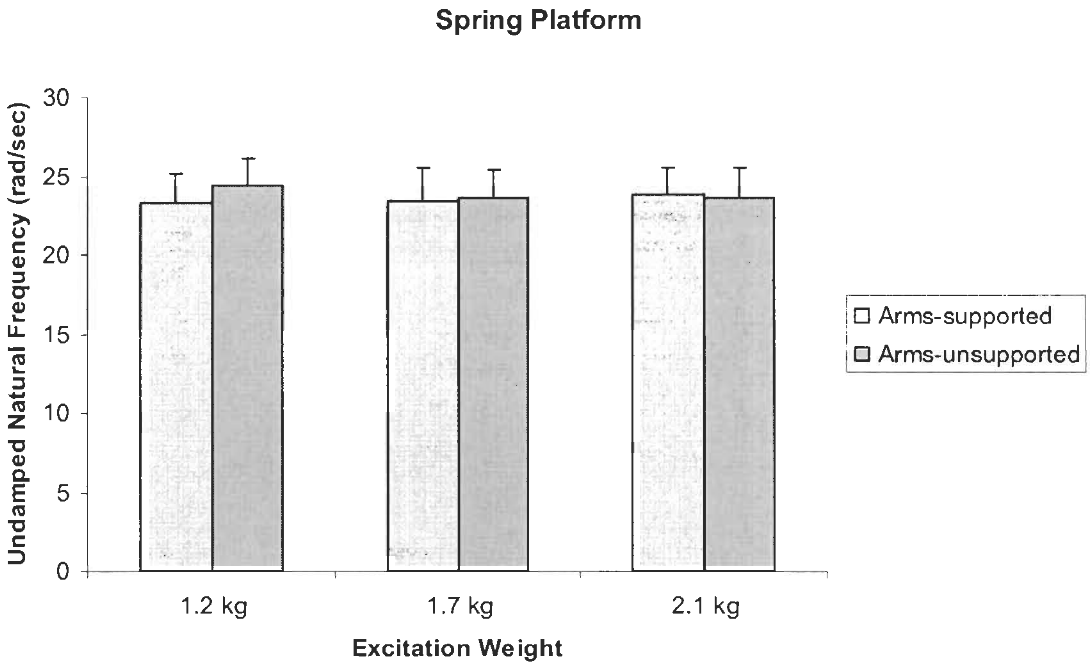
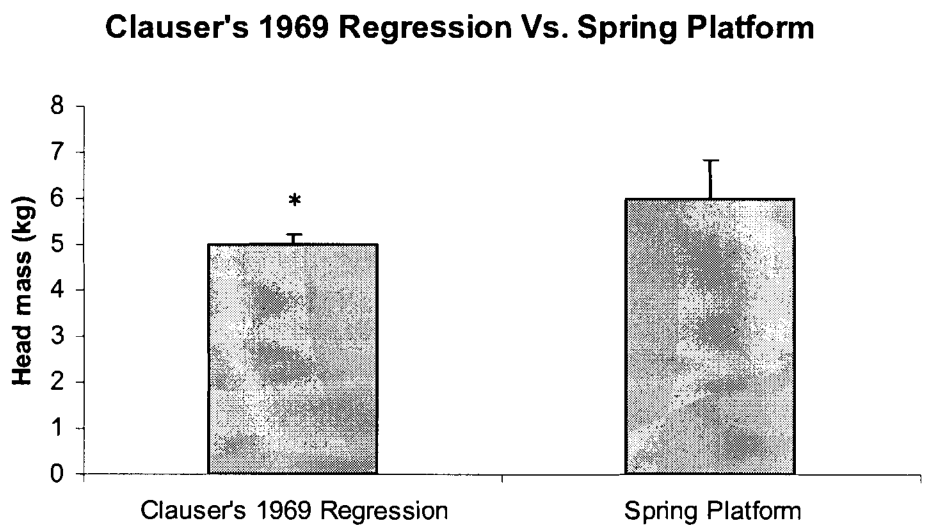


Figure 8: Composite score of head mass for Clauser's 1969 regression and the spring platform experiment.



* The mean difference is significant at $p<0.05$

Figure 9: **Effective head mass appears to correlate with head circumference (r = 0.73).**

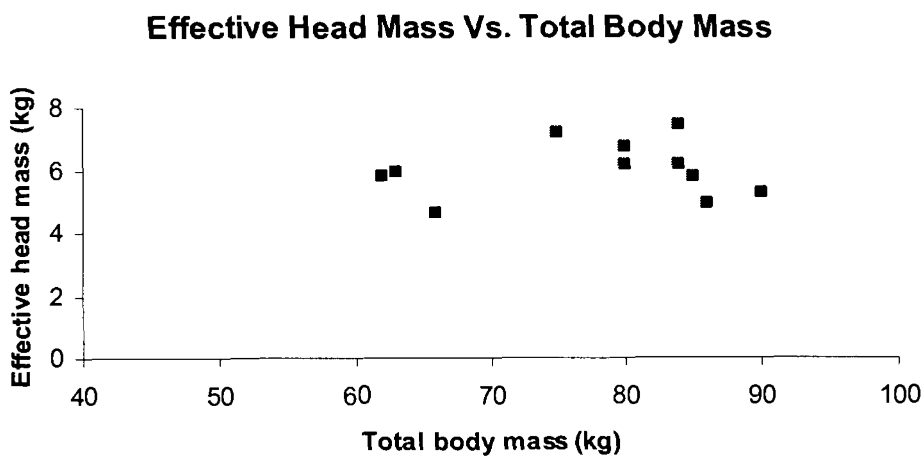
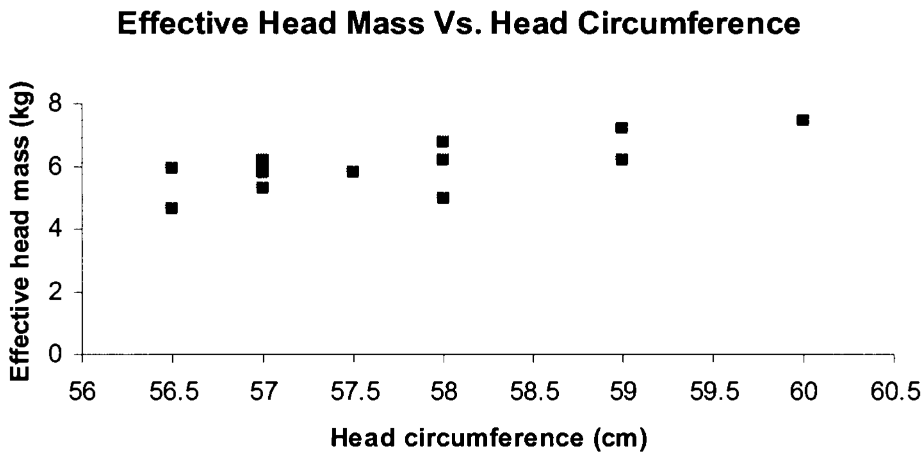


Figure 10: Effective head mass does not appear to correlate with total body mass (r = 0.13).



Chapter 3: Effect of Mouthguards on the Transmission of Force across the Human Jaw

3.1 Abstract

Conclusion: The findings suggest that wearing any mouthguard is better than wearing none at all in reducing the transmission of force from the jaw to the cranium. **Purpose:** This study assessed how the impact response (stiffness and damping values) of the jaw and neck combined are affected by mouthguards with varying material and manufacturing properties, in controlled conditions involving specified upper limb orientation and applied loading. Stiffness and damping values are relevant because they are the major determinants of the impact force, experienced by the individual, and hence injury risk. It is the first study to show experimentally the quantitative effects of mouthguard *in situ*. **Method:** To obtain measures of the head system's free vibration response, a step change in vertical force was applied at the chin. The participants' (n=12) heads were perturbed by removing an excitation weight resting on their heads. Three different predetermined excitation weights were used: 11.5N (1.2 kg), 16.5N (1.7 kg), and 20.8N (2.1 kg). Data were collected with a load cell placed in series with a chin platform. To determine the effect of upper limb orientation, two arm positions were used: arms supported and arms unsupported. Three commercially available boil-and-bite mouthguards were tested to determine the force attenuation characteristics of each. **Results:** All the mouthguards lowered the system stiffness as compared to the no

mouthguard condition ($p < .001$). There was no observed effect on stiffness between the two limb orientation positions. Excitation weight had an unexpected effect on system stiffness ($p = 0.041$), with increasing weight leading to increased stiffness.

3.2 Introduction

Sport-related head injury is an important public health problem (Bailes and Cantu, 2001; Powell and Barber-Foss, 1999; Powell, 2001; Ruchinskas, et al., 1997; Thurman, et al., 1998). In the latest year for which data are available, approximately 306,000 brain injuries were attributed in 1997 to sports or other physical activity in the United States that year, accounting for approximately 20% of all brain injury (Thurman, et al., 1998). Rates of traumatic brain injuries were highest in the 15-24 year age groups. Since the quality of life of the injured athletes is potentially jeopardized due to prolonged symptoms of concussion, sport-related brain injury is a concern worth addressing.

To assess sport-specific risk factors, the incidence rates of head injury in various sports have been collected. Sports that have been identified to have high incidences of head injuries include: boxing, football, soccer, rugby, ice hockey, wrestling, and basketball (Bailes and Cantu, 2001; Powell and Barber-Foss, 1999; Powell, 2001; Ruchinskas, et al., 1997; Thurman, et al., 1998). It has been shown that football accounts for 63% of all reported head injury in high school sports in the United States (Powell and Barber-Foss, 1999). In a study conducted in British Columbia, Canada, an incidence rate of 5.95 concussive injuries per 1000 player/game hours has been reported in Junior A level hockey (Goodman, et al., 2001). The potential seriousness of head injury appearing in sports is increasingly being recognized through continuing research in the area of sport-related head injuries.

It is generally accepted that a mouthguard's primary function is to provide injury prevention to teeth. Recently, numerous investigators have suggested that mouthguards may have the potential to reduce the occurrence of concussion in athletes (Benson, et al., 2002; Biasca, et al., 2002; McCrory, 2001; McCrory, 2001).

It has been hypothesized that the protective nature of mouthguards against concussion lies in the ability of the mouthguard to absorb energy when a force is transmitted vertically across the occlusal surface of the teeth. From studies that investigated various mouthguard materials and designs, it was found that the energy absorption is directly correlated to the thickness of the material used (Bemelmanns and Pfeiffer, 2001; Hoffmann, et al., 1999; Tran, et al., 2001). These studies, however, were carried out *in vitro*. Intuitively, the most protection should be provided by thick shock absorptive materials between the occlusal surfaces of the teeth. To date, however, the testing of mouthguards *in situ* has not been reported.

Stiffness and damping are two mechanical variables deemed most important, because along with effective mass and impact energy these two variables govern impact force and injury risk. The purpose of this study was to assess how the impact response (stiffness and damping values) of the jaw and neck were affected by the use of mouthguards. A secondary objective was to compare the stiffness and damping of different boil-and-bite mouthguards, with varying material and manufacturing properties, in controlled conditions involving specific upper limb orientations, and excitation weight. The purpose of the excitation weights was to determine the stiffness linearity. Ideally, the different excitation weights used to perturb the head will not have an effect on the calculated stiffness and damping values. Once stiffness linearity is established, any change in system stiffness can be attributed to either the mouthguard or the limb orientation conditions. It was hypothesized that any mouthguard will be better in reducing system stiffness than having none at all. In addition, it was expected that system stiffness will decrease with more absorptive composition of the mouthguards.

3.3 Methods

3.3.1 Subjects

12 male subjects between the ages of 19 to 28 participated in this study. All subjects were healthy, had their full set of teeth (± third molars), and no mechanical (orthodontic) appliances. Prior to the experiment a Physical Activity Readiness Questionnaire (PAR-Q) was administered to assess the general health status of the participants, and an informed consent was obtained. Ancillary measures on the participant's weight and head circumference were taken. These two ancillary measures were used to approximate the mass of the head using Clauser's 1969 regression formula (Marfell-Jones, 1984). The participants were each fitted with three different boil-and-bite mouthguards (model 1: Ez Guard [Minnesota, USA]; Model 2: Shock Doctor [Minnesota, USA]; Model 3: Brain Pad [Pennsylvania, USA]). The mouthguard were all fitted according to each manufacturer's instructions.

3.3.2 Experimental Protocol

A specific experimental protocol was developed in order to obtain measures of the head system's free vibration response to a step change in vertical force applied at the chin. Participants sat erect on a stool and placed their chins on a chin platform. Plasticine was glued on to the chin platform and formed to fit the participants' chin. The chin platform attached directly over a "pancake" load cell (Sensotec® Tension/Compression load cell Model 41), which was used to record the changes in force across the load cell. The load cell was secured on the bottom to a tabletop. To allow each subject to sit uniformly, the height of the table was adjusted with the use of a hydraulic pump. An excitation weight was placed on the subjects' heads, and once the signal tracing had stabilized for five seconds, the excitation weight was quickly removed and data were collected for five seconds using DATAQ™ at a sample rate of 488 Hz. The excitation weight had a concave plasticine base that was used to stabilize it unto the participants' heads. The excitation weight was removed by using a counter-weight attached to it through a pulley system secured to the ceiling (Figure 11). Predetermined excitation weights were: 11.5N (1.2 kg), 16.5N (1.7 kg), and 20.8N (2.1 kg) were used. Prior to the removal of the excitation weight the subjects were instructed to relax their neck while the data was being collected.

Each participant underwent four mouthguard test conditions plus one control condition. For the control (no mouthguard) the subjects were asked to lightly close their mouths so that the maxillary and mandibular teeth were in complete occlusion without clenching the jaw. In the remaining three conditions three different types of mouthguards were used. Each of the four conditions were in turn performed with the three predetermined excitation weights.

To determine the effect of arm orientation on system stiffness and damping values, two arm positions were used: arms-supported and arms-unsupported. In the arms-supported position, the subjects were instructed to place their arms on the table and use them to support their upper body weight. Although the extent to which the participants supported their torso was not measured, we were more interested in comparing the effects of upper limb positions. In the arms-unsupported position, the subjects placed their hands on their lap with five trials per condition. In total, each subject underwent twenty-four different head perturbation conditions. Counter-balancing of participants was achieved by having each of the twelve participants start at a different condition, for both arms-supported and arms-unsupported conditions.

3.3.3 Human Head and Mouthguard-Head Models

A head perturbation experiment was designed to measure the dynamic response of the head to an applied step change in vertical force. Conceptually, the simplest model capable of simulating the dynamic response of the head during the experiment consists of a single effective mass (m) attached to a parallel spring-damper elements (k and b) (Figure 12). The effective mass is located at the head and the spring-damper element is mainly located in the soft tissues surrounding the jaw and neck. With a mouthguard in place it was expected that the mouthguard's spring-damper elements would contribute to the measured parameters. Therefore, in experiments with a mouthguard the measured k and b values are contributed by both soft tissues and the mouthguard. Any differences in these values are then solely due to the type of mouthguard being tested.

3.3.4 Parameter Identification

In all experiments, the damped natural frequency ω_d and damping ratio ξ were measured (Figure 13). The damped natural frequency was equal to

(1) $$\omega_d = \frac{2\pi}{T_d}$$

where T_d was the damped natural period equal to twice the time interval between the first force maximum and the first force minimum. The damping ratio was approximated by

(2) $$\xi = \frac{\delta}{\sqrt{\pi^2 + \delta^2}}$$

where δ was the logarithmic decrement,

(3) $$\delta = \ln\left(\frac{F_1}{F_2}\right),$$

F1 was the difference between the first force maximum and the end load, and F2 was the difference between the end load and the first force minimum (see Figure 13). The un damped natural frequency was ω_n given by

(4) $$\omega_n = \frac{\omega_d}{\sqrt{1-\zeta^2}}.$$

Finally, stiffness *k* and damping *b* values were calculated as

(5) $$k = \omega_n^2 m$$

and

(6) $$b = 2\xi\omega_n m.$$

Where the effective mass *m*, the weight of the head, was required, it was obtained using values predicted using the spring-platform technique (see Appendix A).

3.3.5 Data Analysis

Following computation of both *k* and *b* for each trial, the median *k* and *b* values were computed. The median value, chosen as a relevant outcome variable over the mean due to limited number of trials in each condition, then served as the dependent variable in the inferential analysis. A 2 (arm position) x 3 (excitation weight) x 4 (mouthguard) repeated measures ANOVA was used. When a significant F value was detected for the Excitation Weight main effect, a Bonferroni corrected pairwise comparison among estimated marginal means was employed to determine the position of the difference. For the Mouthguard factor, a Helmert contrast was used. Helmert contrasts compare: the first level of the factor with all later levels, the second level with all later levels, and so forth. The first level was No Mouthguard, and the subsequent levels were assigned according to increasing retail price. Significance was set at $p < .05$ and all data are presented as mean ± SD.

3.4 Results

3.4.1 Subjects

Ancillary measures (head circumference and total body mass) were collected on all 12 subjects and is presented in Table 2. Using the spring-platform technique the effective mass of the head was determined for each subject, with a mean of 6.01 ± 0.24 kg.

3.4.2 Mouthguard Condition

Table 3 shows the parameter values for each mouthguard condition for each subject. These values were used to calculate system stiffness and damping values, as well as the weight of the head. The plot of the mean calculated system stiffness for each mouthguard is shown in Figure 14. There was a significant difference ($F_{3,44} = 61.27$; $p < .001$) when the No Mouthguard condition (13270 ± 743 N/m) was compared with the mean stiffness of the three mouthguard conditions (9052 ± 529 N/m). In the second contrast, there was a significant difference ($F_{3,44} = 9.06$; $p = 0.012$) when Model 1 (9670 ± 328N/m) was compared with the mean of Model 2 and Model 3 combined (8744 ± 630 N/m). Lastly, there was a significant difference ($F_{3,44} = 4.94$; $p = 0.048$) between Model 2 (9107 ± 722 N/m) and Model 3 (8380 ± 538 N/m).

The plot of the mean calculated system damping values for each mouthguard is shown in Figure 15. In the first contrast, there was a significant difference ($F_{3,44} = 20.38$; $p < .001$) when the No Mouthguard condition (88.2 ± 6.4 N*s/m) was compared with the mean of all three mouthguard conditions (66.3 ± 5.0 N*s/m). In the second contrast, there was a significant difference ($F_{3,44} = 13.48$; $p = 0.0037$) when Model 1 (76.4 ± 5.3 N*s/m) was compared with the mean of Model 2 and Model 3 combined (61.3 ± 4.9 N*s/m). There was no statistically significant difference ($F_{3,44} = 0.56$; $p = 0.47$) between the mean system damping of Model 2 (58.0 ± 4.4 N*s/m) and Model 3 (64.7 ± 5.4 N*s/m).

3.4.3 Excitation Weight

A decrease in calculated stiffness was observed with increasing excitation weight: 14232 ± 4237 N/m, 12373 ± 2404 N/m, and 11588 ± 3142 N/m for 1.1 kg, 1.7 kg, and 2.1 kg, respectively (Figure 16). A significant Excitation Weight main effect ($F_{3,44} = 3.72$; $p = 0.041$) was detected for system stiffness. The difference was located between 1.1 kg and 1.7 kg (mean difference = 1860; 95% CI = 131 to 3589; $p = 0.037$). An

Excitation Weight main effect was not detected for system damping ($F_{3,44} = 1.79$; $p = 0.19$) (Figure 17).

3.4.4 Arm Position

No significant difference was detected when system stiffness of arm supported 12598 ± 4018 N/m was compared with arm unsupported 12863 ± 2466 N/m (mean difference = 264; 95% CI = -2767 to 2239; p = 0.821) (Figure 18). Similarly, no significant difference was detected when system damping of arm supported 97.4 ± 27.5 N*s/m was compared with arm unsupported 97.9 ± 21.6 N*s/m (mean difference = 0.46; 95% CI = -13.45 to 12.54; p = 0.94) (Figure 19). Since no significant arm-position effect was detected the data was collapsed over this condition.

3.5 Discussion

As we had hypothesized, there was a significant reduction in system stiffness between the no mouthguard condition and the mean of all three boil-and-bite mouthguards combined. The introduction of an absorptive material, such as polyethelene, within the occlusal area functions to decrease the system stiffness of the head. Model 1 yielded a higher stiffness when compared to a combination of Model 2 and Model 3. One possible explanation for this difference could be that Model 1 contains a single composite material (polyethelene), whereas Model 2 and Model 3 use a multi composite construction. Both Model 1 and Model 2 mouthguards are fitted on the maxillary arch. In comparison, Model 3 is a bi-maxillary mouthguard, fitted on both the maxillary and mandibular arches.

The pre-formed Model 3 (13mm thickness) was also thicker than the Model 1 (3mm thickness) or Model 2 (8mm thickness) mouthguards. According to other studies, shock absorption is directly related to material thickness (Bemelmanns and Pfeiffer, 2001; Hoffmann, et al., 1999; Tran, et al., 2001); thus Model 3 should provide the greatest protection. In this study, Model 3 had the lowest mean calculated stiffness. However, consideration must be given to the compliance of athletes with respect to the continued use of bulkier mouthguards.

Similar to system stiffness, there was also a significant reduction in system damping from the no mouthguard condition to the mean of all three boil-and-bite mouthguards combined. Damping refers to the ability of the system to dissipate energy, usually as heat. Unlike stiffness, damping is a property of the system whose ability to provide a protective function is less well understood. Whereas stiffness is correlated with impact force— a decrease in stiffness will decrease injury risk, the effect of damping is yet unknown.

Materials used in the fabrication of mouthguards appear to act primarily as a spring, an energy absorber, and not much like a damper, an energy dissipator. An energy absorber stores the energy and returns it back, whereas an energy dissipator dissipates the energy usually in the form of heat. The mouthguard, *in situ*, causes a decrease in the system's effective damping constant b, and a decrease in the effective stiffness k. Stiffness and damping, along with effective mass and impact velocity, determines impact force and injury risk. Therefore, both of these changes will reduce peak impact force, for a given impact energy.

Contrary to our expectation, the excitation weight did have a minimal effect on the calculated system stiffness *k*. That is, there was a significant difference between 1.1 kg and 1.7 kg excitation weights. This was most likely due to the fact that the lowest weight used was in the initial "toe region" of the linear system stiffness. However, for our purposes, it was important to start with a light excitation weight given we were perturbing intact human subjects. Biological tissues conform to linear system stiffness after an initial "toe region". While our system also included a mechanical insert (mouthguard) this would not contribute to the non-linearity of the "toe region" since mechanical system are generally linear throughout their full region of operation, until failure. Unlike system stiffness, damping appeared to be stable across the three excitation weights.

Two arm position conditions were used to investigate the effect of the upper limb and torso on the calculated *k* and *b* values. Neither the system stiffness or damping values were different between arm-supported and arm-unsupported positions. Since no difference was detected, the posture of the torso and upper limbs do not appear to affect system stiffness or damping. Thus, in this particular experimental setup, the measured values most probably reflect the stiffness and damping characteristic of the head and neck system.

3.6 Conclusion

In conclusion, we found that there was a reduction in system stiffness and damping values when a boil-and-bite mouthguard was introduced into the perturbed head and neck system. Moreover, there was a difference in the change of these values determined by the brand of the mouthguard.

When determining which mouthguard to purchase, one should consider the protection it provides, as well as the likelihood the athlete will wear the particular mouthguard on a continual basis. Even though a particular mouthguard may provide the most protection, its purpose is defeated if there is low compliance. It is hoped that this research will stimulate additional exploration into the potential benefits of mouthguards against concussion.

3.7 References

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Table 2: Subject characteristics. Values are reported as means ± SD.

Age (yrs)	Forehead girth (cm)	Body mass (kg)	Calculated head mass (kg)*
23.7 ± 1.1	57.79 ± 0.32	78.3 ± 2.8	6.01 ± 0.24

* Values predicted using spring-platform technique.

Table 3: Parameter values and calculated stiffness and damping values for each subject.

Subject	Mouthguard	k (N/m)	b (N*s/m)	ω_n (rad/sec)	Damping Ratio
1	No Mouthguard	11481	102.6	46.8	0.206
	Model 1	8683	66.6	40.7	0.164
	Model 2	8719	55.3	40.6	0.131
	Model 3	7540	64.0	37.7	0.166
2	No Mouthguard	12169	78.6	49.3	0.166
	Model 1	11025	69.3	46.5	0.148
	Model 2	9689	63.2	43.9	0.161
	Model 3	9990	69.1	44.7	0.161
3	No Mouthguard	16129	85.3	53.8	0.162
	Model 1	13674	76.0	49.7	0.144
	Model 2	9784	77.1	43.0	0.177
	Model 3	9412	80.8	42.1	0.186
4	No Mouthguard	10205	95.4	44.0	0.207
	Model 1	8175	75.1	39.9	0.189
	Model 2	6919	58.5	36.3	0.154
	Model 3	6409	57.7	34.7	0.165
5	No Mouthguard	11733	73.7	49.2	0.164
	Model 1	9672	77.6	44.9	0.179
	Model 2	7639	41.0	40.2	0.104
	Model 3	7200	47.4	38.9	0.122
6	No Mouthguard	12044	98.2	49.3	0.196
	Model 1	9857	84.4	43.8	0.187
	Model 2	8455	76.5	40.6	0.183
	Model 3	8527	74.1	40.6	0.174
7	No Mouthguard	17207	85.8	54.5	0.157
	Model 1	10296	105.9	43.1	0.216
	Model 2	11748	84.6	46.6	0.169
	Model 3	9773	82.7	42.0	0.185
8	No Mouthguard	14449	136.9	52.9	0.249
	Model 1	9673	101.7	43.3	0.234
	Model 2	8834	87.0	41.6	0.200
	Model 3	9850	110.0	43.9	0.242
9	No Mouthguard	13783	90.9	51.1	0.171
	Model 1	11535	99.0	46.9	0.197
	Model 2	11153	82.2	46.2	0.173
	Model 3	9553	85.9	42.8	0.195
10	No Mouthguard	20311	88.2	62.0	0.153
	Model 1	12557	83.2	47.9	0.166
	Model 2	15109	81.2	52.8	0.151
	Model 3	12567	53.4	47.8	0.122
11	No Mouthguard	14567	100.8	52.9	0.196
	Model 1	8762	81.1	41.9	0.217
	Model 2	6038	59.2	35.6	0.176
	Model 3	5787	68.8	34.6	0.201
12	No Mouthguard	17477	145.7	54.6	0.236
	Model 1	11344	108.9	43.0	0.236
	Model 2	8354	79.4	36.1	0.202
	Model 3	8735	88.7	37.0	0.225

Subject	Mouthguard	k (N/m)	b (N*s/m)	ω_n (rad/sec)	Damping Ratio
$\bar{x}$ ± SD	No Mouthguard	14791±3063	98.5±21.8	51.7±4.6	0.189±0.032
	Model 1	10555±1710	85.7±14.5	44.3±3.0	0.190±0.031
	Model 2	9403±2671	70.4±14.5	42.0±5.0	0.165±0.028
	Model 3	8781±1961	73.6±17.4	40.6±4.1	0.179±0.036

Figure 11: Setup of experiment. When the counter-weight is released the excitation weight is lifted off the head.

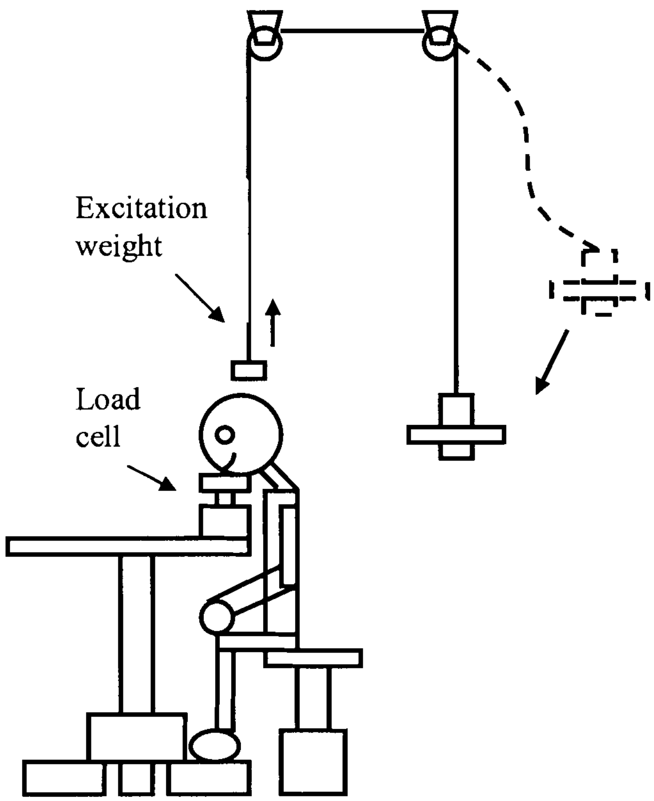


Figure 12: Schematic of the under-damped spring-dashpot system with one degree of freedom.

The model on the upper left shows individual k and b of the jaw (j) and neck (n). This model can be reduced to a single spring-damper-mass system (shown on the upper right). With the addition of a mouthguard the same model can be used (shown on the lower right).

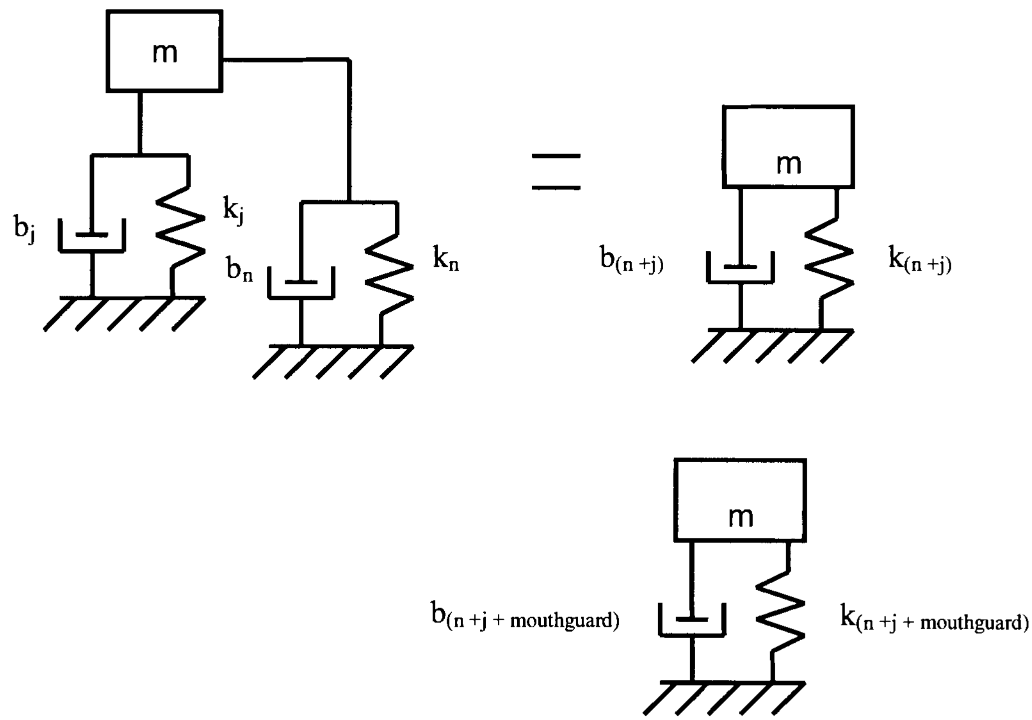


Figure 13: One trial output from DATAQ

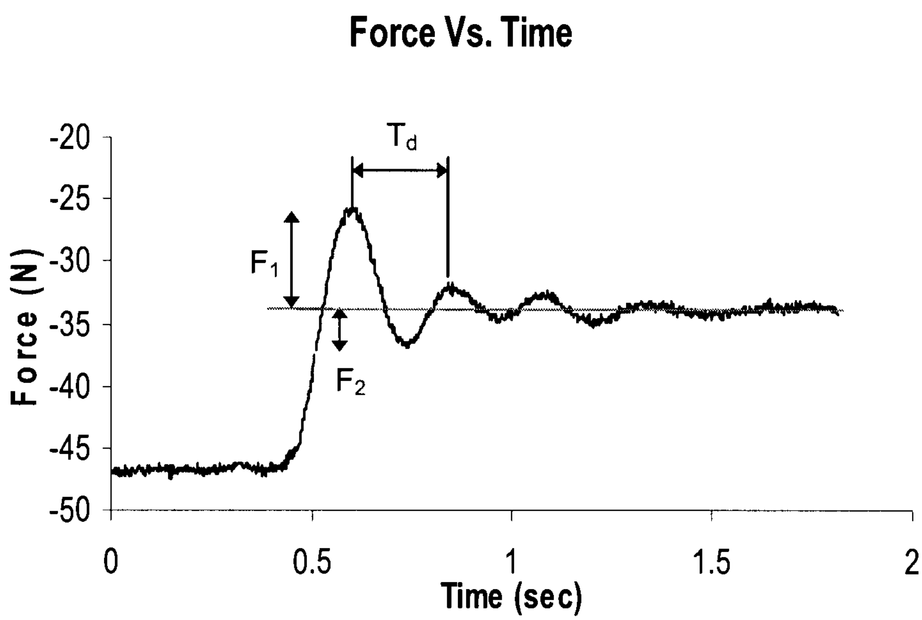
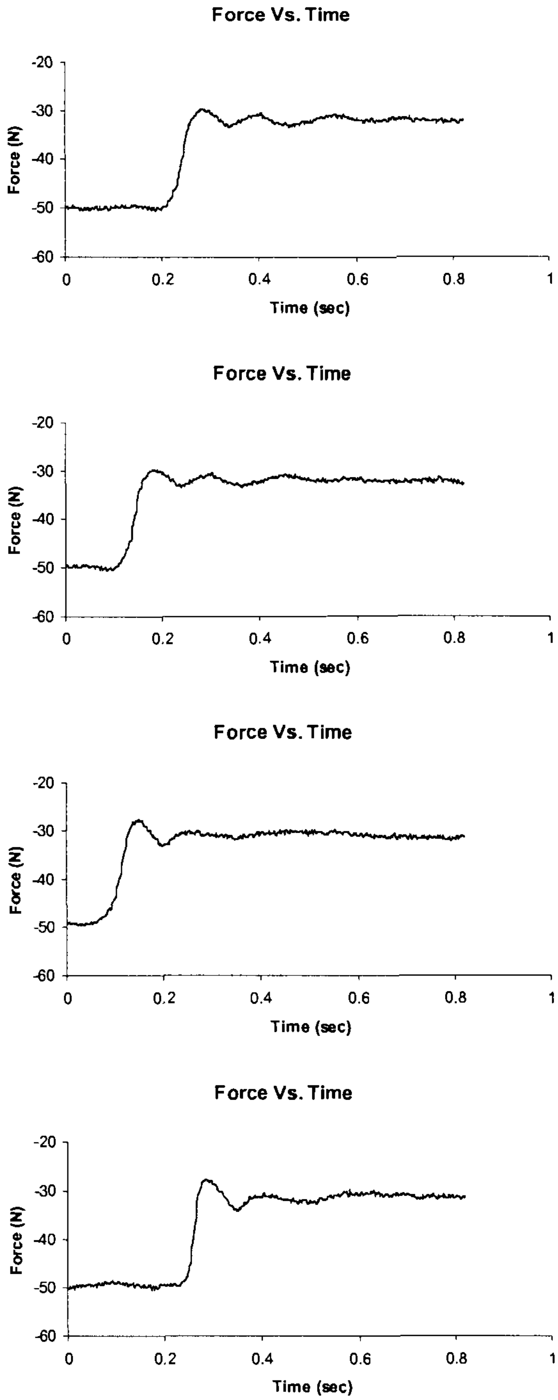


Figure 14: Representative graphs of the four mouthguard conditions.



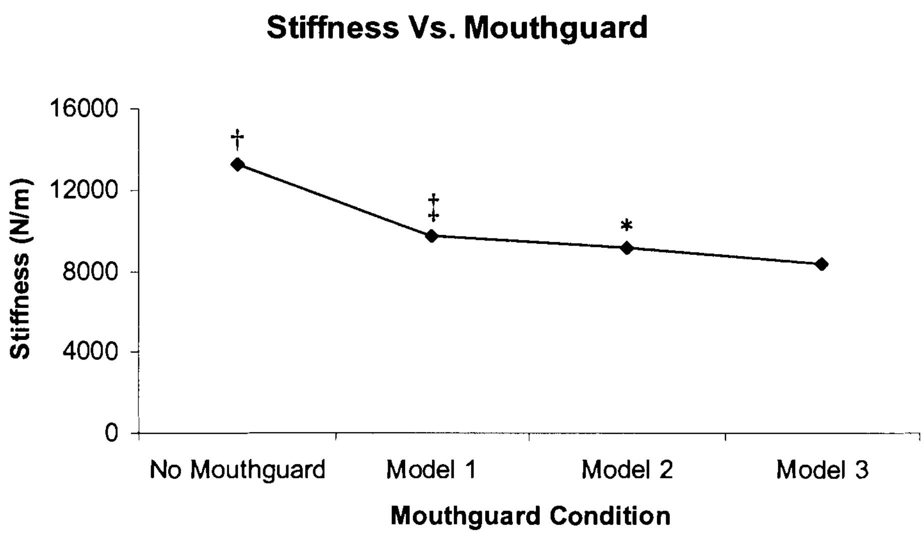
No Mouthguard

Model 1

Model 2

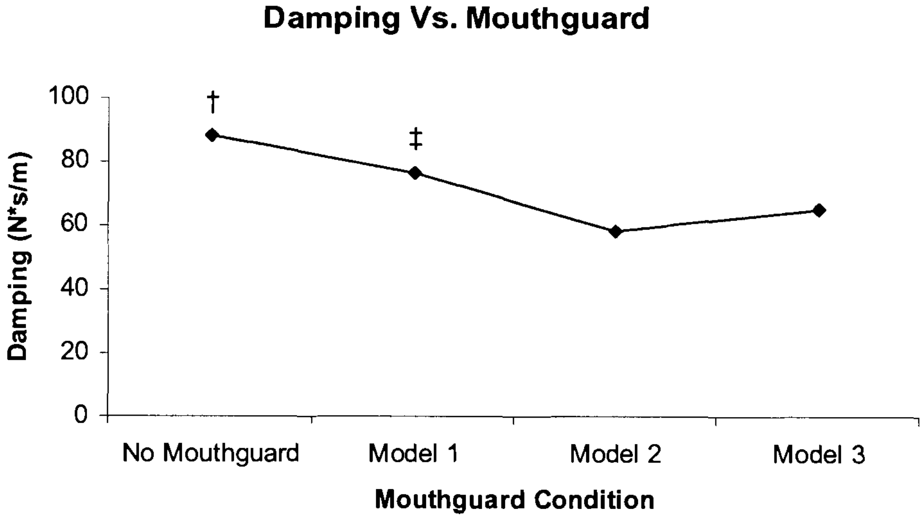
Model 3

Figure 15: Composite score of calculated stiffness values for different mouthguard conditions (n=12).



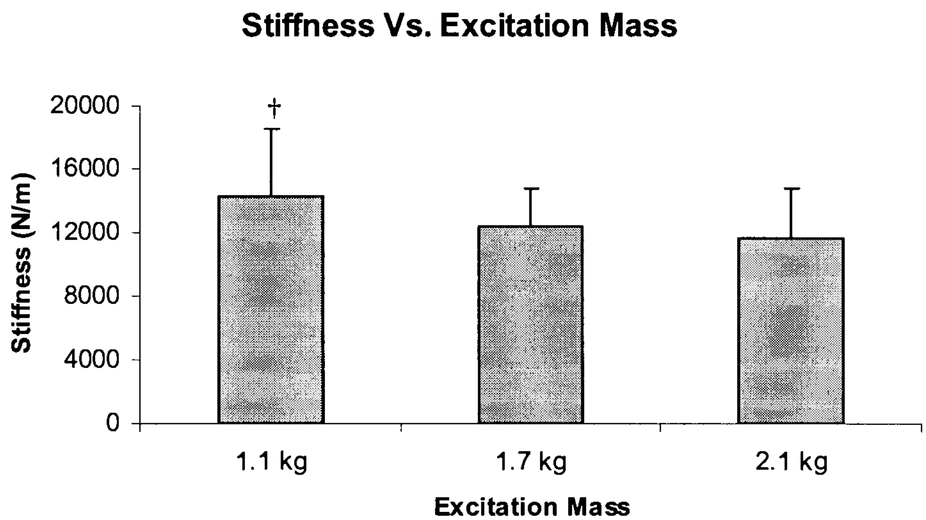
† Significant greater stiffness for No Mouthguard condition compared with mean of all three mouthguards ($p < .001$)
‡ Significant greater stiffness for Model 1 compared with mean of Model 2 and Model 3 combined ($p = 0.012$)
* Significant greater stiffness for Model 2 compared with Model 3 ($p = 0.048$)

Figure 16: Composite score of calculated damping values for different mouthguard conditions (n=12).



† Significant greater damping for No Mouthguard condition compared with mean of all three mouthguards ($p < .001$)
‡ Significant greater damping for Model 1 compared with mean of Model 2 and Model 3 combined ($p = 0.0037$)

Figure 17: Composite score of calculated stiffness values for different excitation weight conditions (n=12).



† Significant greater stiffness for 1.1 kg compared to 1.7 kg (p = 0.037)

Figure 18: Composite score of calculated stiffness values for different excitation weight conditions (n=12).

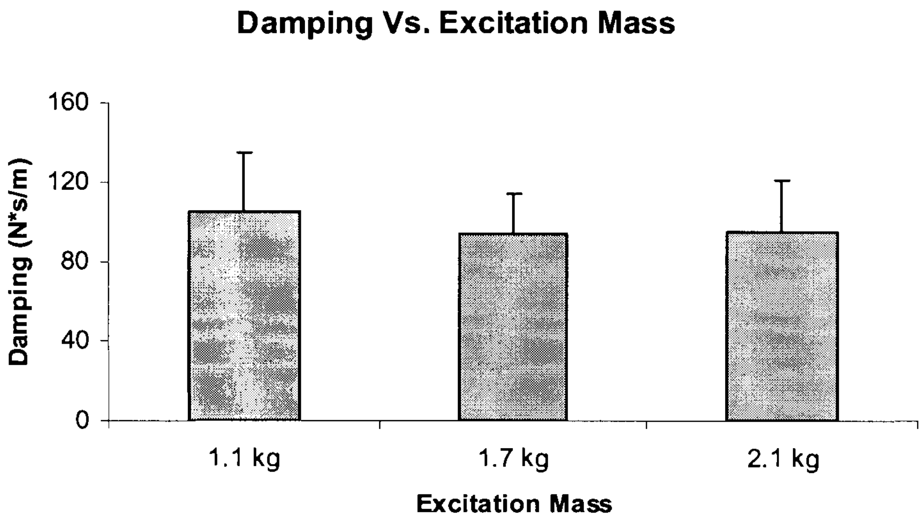


Figure 19: Composite score of calculated stiffness values for different arm position conditions (n=12).

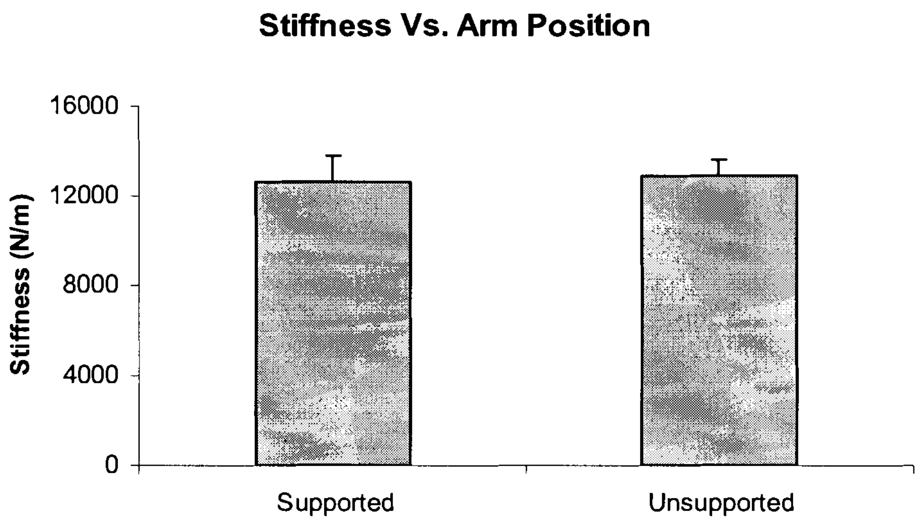
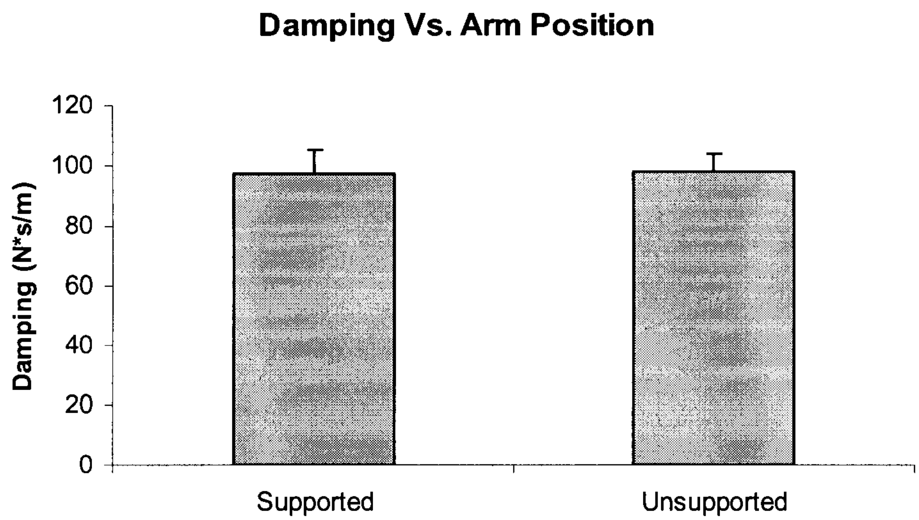


Figure 20: **Composite score of calculated damping values for different arm position conditions (n=12).**



Chapter 4: General Discussion and Conclusion

The ability of mouthguards to reduce the occurrence of concussions in athletes is not well understood and is largely inferred from dental trauma studies. However, the present study is the first to provide *in situ* data on the quantitative effects of mouthguards. This is significant when considering the numerous mouthguard advertisements that promote protection against concussion. The decrease in system stiffness observed in this study may have implications in evaluating different mouthguards.

As it was hypothesized, the jaw and head system stiffness was decreased with the introduction of a mouthguard. The significant decrease in system stiffness indicates that mouthguards act primarily as a spring, an energy absorber. Therefore, a properly fitted mouthguard is expected to reduce peak impact force, for a given impact energy, and decrease injury risk. This is further supported by previous studies which have shown that peak impact force is dependent on the material used to fabricate mouthguards and on its thickness (Bemelmanns and Pfeiffer, 2001; Hoffmann, et al., 1999; Tran, et al., 2001). This study is the first to show experimentally that mouthguards may potentially reduce the risk of concussion when the force is directed across the mandible to the cranium.

Another objective of this study was to compare different mouthguards using the head perturbation technique developed in this study. The effects of three different commercially available mouthguards on system stiffness and damping were examined.

Of relevance to injury risk prevention, system stiffness is correlated with impact force – a decrease in system stiffness will decrease injury risk. The three mouthguards differed in their abilities to decrease system stiffness. The Brain Pad mouthguard showed the greatest decrease in system stiffness, followed by the Shock Doctor and EZ Guard mouthguards, respectively. As it was shown in previous studies, there was direct relationship between shock absorption and material thickness (Bemelmanns and Pfeiffer, 2001; Hoffmann, et al., 1999; Tran, et al., 2001). The Brain Pad mouthguard was the thickest, covering both maxillary and mandibular arches. Whereas, the Shock Doctor and EZ Guard only covered the maxillary arch, with the EZ Guard being the thinner of the two. Mouthguards undoubtedly alter system properties, the results indicate that *in situ* thicker mouthguards should provide better protection.

As a secondary objective the effects of excitation weights and arm positions were investigated. For ethical and safety issues, it was important to start with light excitation mass. Of the three excitation masses (1.1kg, 1.7kg and 2.1kg) used in this study, the 1.7kg excitation mass marks the beginning of the linear stiffness region of the head and neck system combined. Above 1.7kg the calculated system stiffness should be the same irrespective of the excitation weight and any changes seen in the stiffness would be due to the experimental parameters, such as different mouthguards.

No difference in system stiffness and damping were detected whether the arms were supported on the table or not. When using the head perturbation technique the calculated stiffness and damping values reflect the head and neck system.

The head mass is required to calculate stiffness and damping values of the head and neck system. Traditionally, segmental masses are determined using anthropometric measurements, such Clauser's 1969 regression (Marfell-Jones, 1984; Nigg andLiu, 1999). The benefit of this method is that the measurements are easily obtained at minimal cost. However it is inappropriate to use this method when stimulating dynamic conditions since it only estimates static masses. When studying impact situations it is more appropriate to use the effective mass because it simulates realistic conditions (Liu and Nigg, 2000). For this reason, some investigators utilize perturbation experiments to determine segmental masses (Lin, et al., 2001; Robinovitch, et al., 1997). As there is yet a perturbation experiment developed to estimate the head mass, one had to be built and validated.

Perturbation experiments are based on the fact that biological tissues can be considered as a mechanical system with masses connected by springs and dampers. As expected, the perturbation of the head produced the response of an under-damped spring-dashpot system with one degree of freedom. The head perturbation experiment developed in this study properly measures the biological response of the head and neck system to a perturbation.

The effective mass of the head obtained from the perturbation technique was compared with Clauser's 1969 regression. The effective mass from the perturbation experiment was higher than the static mass from Clauser's method. This indicates the higher mass observed in the perturbation method are due to mechanical factors that cannot be measured through anthropometric methods. Such factors could include muscle tone, ligaments and other soft tissues.

This was the first study to develop and experimental method to estimate the effective mass of the head from an intact human being. In addition, with minor changes to testing protocols the effects of mouthguards *in situ* on the head and neck system properties were examined for the first time. Clinically, this method can be used to stimulate realistic impact situations to the head and potentially screen protective appliances, such as mouthguards and helmets.

4.1 References

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Appendix A

The un-damped natural frequency ω_n was calculated for both conventional and spring-platform condition. In the conventional condition k and m contributed to ω_{n1}. Whereas, in the spring platform condition *k*, *m*, and k_s contributed to ω_{n2}. Therefore, ω_{n1} and ω_{n2} were equal to

(1) $$\omega_{n1} = \sqrt{\frac{k}{m}}$$

and

(2) $$\omega_{n2} = \sqrt{\frac{\left(\frac{k * k_s}{k + k_s}\right)}{m}}.$$

The spring platform stiffness k_s was measured from free vibration tests and was equal to

(3) $$k_s = \omega_n * 1.12kg.$$

By combining Eq. 1, 2, and 3, the effective mass *m* was given by

(4) $$m = k_s\left(\frac{1}{\omega_{n2}^{\ 2}} - \frac{1}{\omega_{n1}^{\ 2}}\right).$$