THE BIOMECHANICS OF HEAD TRAUMA AND THE DEVELOPMENT OF THE MODERN HELMET. HOW FAR HAVE WE REALLY COME?

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ABSTRACT

This paper reviews the development of protective headgear over the past 50 years and attempts to put in perspective the role of biomechanics in that process. Historically, basic engineering concepts of energy absorption and load distribution have been applied to helmet design and performance specifications but little in the way of biomechanics of head injury per se has been utilized. The lack of progress appears to stem largely from adherence to old fashion test methods which may not properly reflect the circumstances in which most helmeted humans may find themselves. The biofidelity of the test device, the nature of the failure criteria, as well as the manner by which the movement of the test helmet is constrained, are all issues that bear on the application of biomechanical fundamentals. The paper concludes with suggestions regarding how to better implement what is known about the biomechanics of head injury to helmet design and standards development.

Keywords: BIOMECHANICS, HELMETS

50 YEARS AGO, Juan Fangio of Argentina was the World Champion of motor car racing. When he competed, he wore head protection that has been described as a "pudding bowl". Bill Lomas the World Champion motorcyclist from the UK wore a similar "helmet". Otto Graham, voted the Most Valued Player of the National Football League in the United States in 1955, wore a hard leather hat when he played. Cyclists 50 years ago may have worn what could best be described as a leather hairnet and the occasional ice hockey player may have worn a soft leather cap. Most other sporting endeavors left protection of the participant's head as something of not much concern or interest.

Until the British Standards Institute published the world's first crash helmet standard in 1952 (BS 1869, 1952), what actually constituted a helmet was left pretty much to the discretion of the helmet producer. Performance standards like this were the first real opportunity to invoke human tolerance considerations, i.e. biomechanics, for it was such standards that began to require that a helmet be defined in terms of how it should function rather than how it was designed or manufactured. Helmet standards now required some knowledge of human head injury threshold for how else could they stipulate a failure level?

Helmets of course have been around for thousands of years and it is self evident that their primary function was to reduce the likelihood of head injury in combat (Tenner, 2003). The sports helmet as we know it today has its origins not in the medieval battle field, but in the development of the motorcycle and the airplane.

Our basic understanding of the biomechanics of head injury lies not in the sophisticated computer models of today but in the pioneering work of physicians, physicists and engineers in Europe and America. This paper will not attempt to summarize the wealth of literature dealing with the biomechanics of head injury as that has been done well by others – recently and most notably by McLean and Anderson (1997). Nor will this paper attempt to summarize all the developments in the field of head protection. Instead, we will try to review the progress of helmet design and performance in the context of significant observations from the biomechanics arena.

At the outset, it is clear that a broad understanding of what a helmet should do preceded any clear understanding of head injury mechanisms. The basic principals have always been to provide a hard outer shell to ward off external agents and padding to help cushion the blow. That the shell distributed the applied force and thereby reduced localization of loading, and thus the propensity for skull fracture, may not have occurred to early inventors. That padding served to absorb impact energy thereby reducing the inertial loading on the head and thence reducing acceleration-induced injuries was also not likely given much thought.

Today, there is probably little disagreement that the fundamental objective of good helmet design is to distribute impact loading over as large an area of the head as possible and to reduce the total force on the wearer's head as much as possible.

The main way by which biomechanics has influenced helmet design is not so much in our understanding of different injury mechanisms, but rather in a better appreciation of the biophysical characteristics of the head and the development of kinematic head injury assessment functions. This insight has provided better ways to test the impact capabilities of a helmet without first placing it on a human being and a means to judge how well one might expect it to work in actual use.

HEAD INJURY MECHANISMS.

Not unlike the structural failure of any inanimate object, injuries to the head are nearly always caused by excessive movement of one part relative to another.

Injury to the brain can occur if any part of it is distorted, stretched, or compressed, or if it is torn away from the interior of the skull. Blood vessels rupture if they are stretched too much. An impact to the head can cause the skull to deform and, even if it does not fracture, the underlying brain tissue can be injured as it distorts under the influence of the deforming skull.

Even if the skull does not bend significantly but the head as a whole is caused to move violently, distortion of the brain within the skull will occur. This typically leads to generalized diffuse injury such as concussion, and in the extreme, coma.

Separating, bending, and stretching are merely descriptors of somewhat different kinds of movement. A helmet, by absorbing some of the impact energy reduces the amount of energy transferred to the head. It thereby can reduce the induced relative movement between parts of the head, and thus the probability and/or severity of head injury.

Early research didn't focus too much on what was going on inside the head when it was struck but relied largely on observation of the overall response of the "victim" by looking at how the head moved dynamically and trying to relate this to the type or severity of brain injury. This field of research became to be broadly known as the biomechanics of head trauma

In the beginning, our understanding of the biomechanics of brain injury was, well, unclear. And to a great extent it remains that way. As Lombard and coworkers put it in 1951 (Lombard et al, 1951),

"There is a bit of confusion in the literature concerning the mechanisms of mechanical injury to the brain."

Thirty years later Goldsmith (1981) observed that;

"The state of knowledge concerning trauma of the human head is so scant that the community cannot agree on new and improved criteria though it is generally admitted that present designations are not satisfactory."

Beginning in the early 1940s, in what were the first of several attempts to try to relate external loading to brain injury, Gurdjian, Webster and Lissner at Wayne State University

(Gurdjian and Webster, 1943) (Gurdjian and Lissner, 1944), impacted living dog's heads. Perhaps not surprisingly now, they observed that the harder the dogs were struck, the more likely or the more serious a head injury would be. In later experiments, they applied jets of air directly to animals' exposed brain. They noted a correlation between the force of the air and the duration of exposure to the blast and the severity of concussive effects.

A few years later, Lissner and co workers continued this line of enquiry with a different model (Lissner, Lebow and Evans 1960). They dropped cadavers onto their heads from different heights onto a flat steel plate until they could produce a skull fracture. Not because they were interested in seeing what it would take to "produce a skull fracture", because they were more interested in finding out what it what it would take to produce a concussion. In these experiments they monitored the intracranial pressure during impact as well as the cadaver head's acceleration. By examining acceleration and relating it to those tests when a fracture occurred, they devised a crude relationship between the probability of concussion, the average head acceleration in Gs and the time duration of the impact. That this should relate to brain injury, particularly concussion, was imbedded in their hypothesis that nearly everyone who sustains a skull fracture sustains a concussion. An adequate description of the head response, i.e., its movement, was felt to be contained entirely within the linear acceleration-time history during the impact. By combining all this data, along with other estimates from accidental free falls, they developed what became and what is still referred to as the Wayne State Concussion Tolerance curve (Gurdjian et al, 1960). This function attempted to relate the level of tolerable linear acceleration to how long that acceleration lasted. One of the many problems with this approach is that most people who sustain a concussion, or many of the other kinds of brain injury, do so without having their skulls fractured. Another problem that followed and that has lingered for years was the idea that brain injury was caused by linear acceleration.

By 1953, the helmet industry was coming to grips with some of these biomechanical concepts as evidenced in the following text from an American football helmet patent (US2,634,415). Something of an oversimplification perhaps, but the inventors recognized that acceleration of the head can be injurious even if the skull does not fracture.

"A head jolt properly may be defined as a sudden and/or severe change in the direction in which the head is moving, or the velocity with which it is moving, or both. The avoidance of sharp and/or severe head jolts is of vital importance. The human brain 'f1oats' in the skull much as the yolk of an egg has floating suspension in its associated egg white. A sharp or severe jolt can rupture an egg yoke without fracturing the egg shell. Similarly, a sharp or severe jolt can cause fatal injury to a human brain without fracturing the skull which houses it, and often with only minor, if any evidence of injury at the outside of the head. There have been many such fatal injuries in the playing of football."

Previously, on the "other side of the pond", a research physicist in the Department of Surgery at Oxford University (Holbourn 1943) had hypothesized that the predominant cause of brain injury was not linear acceleration at all but rather was due to rotation of the head. Holbourn maintained that you could disturb/distort/disrupt the contents of the skull much more readily by rotating the head than by accelerating it linearly - the same way you could with say, a bowl of soup, if you spun the bowl rather than if you just pushed it across the table.[1]

Over 20 years later, researchers in the US, such as Gurdjian, Lissner, Patrick, Hodgson et al, continued to be stuck on the idea that translational(linear) acceleration was the most important mechanism. By the 70s, other US researchers, notably Ommaya and coworkers

[1] Keeping of course a lid on the bowl to more closely simulate a closed head so the "brains" don't spill out.

(Gennarelli et al, 1972) who had subjected live monkeys to linear and angular impact motions, had concluded that *"no convincing evidence has to this date been presented which relates brain injury and concussion to translational motion of the head…"*

Nowadays, it is generally acknowledged that deformation of the brain and associated injury might be understood only by knowing the full three dimensional history of the head's motion following impact. Today internal damage to the brain as a result of head impact is being studied with complex mathematical models that can actually predict the amount of deformation that occurs to the various parts of the brain tissue for different types of impact (King et al, 2003). Some investigators have even gone so far as to put mathematical helmets on the mathematical heads to see what the mathematically predicted effects are (Aare et al, 2003). To do so, in addition to modeling the skull, its contents and the helmet's geometric and physical properties, requires the complete characterization of the head motion in time and in all three dimensions including all linear and rotational components.

HELMET DEVELOPMENT

Initially helmets, for whatever sporting application, were no more than leather bonnets. In auto and motorcycle racing, these designs, usually worn with goggles, were borrowed from earlier aviators and served primarily to keep the head warm and the hair in place. In the late 1800s and early 1900s, American football players and the occasional ice hockey player also wore the equivalent of a soft leather hat. Some employed a fleece or felt lining or were padded somewhat with cotton batting.

The concept of a hard shell, dating back to medieval times, tacitly acknowledged that distribution of the force to the head would reduce the probability of skull fracture – now a biomechanical tenet. Or perhaps it was seen as simply a better way to deflect objects from the head. However, no hard shell appeared on these early "helmets".

In American football, concern for injuries dated back to the early 20th century. In 1905, for example, helmets were not worn and there were 18 deaths and 129 serious injuries. Perhaps not surprisingly, later studies determined that most fatalities in football were due to head injury (Halstead et al, 2004). As demands on the leather football helmet design increased, the outer leather was treated to make it hard (in a similar fashion to that developed many years prior for firefighters' headwear). Individual hard leather pieces were usually sewn to a hard fiber material crown section. Initially, they were simply lined with felt, fleece or some other padding but a few years later, with the introduction of a rudimentary inner suspension, something of a breakthrough in football headgear design had occurred. This new device had the capacity to absorb and distribute blows to the head somewhat more effectively than the floppy leather caps previously in use. But there was still a long way to go.

It would not be until the middle of the 20th century that it was recognized that there were at least two types of sporting headgear. One dealing with the one time life threatening blow that could occur in certain sports (e.g. auto racing), the other to deal with repeated low level blows associated with the game, (e.g. American football, ice hockey, etc.).

The need for a "crash helmet" was born shortly after the widespread introduction of the motorcycle in the 1900s. Companies with names like Triumph in England and Harley Davidson in the US arrived to fill a need for this new means of transport that was much cheaper than the automobile. Of course, "boys will be boys", and motorcycles and cars were being driven faster and faster; often in competition. Crashing at speeds much faster than one could run, skate or cycle, introduced the need for "crash" helmets – though that need would not be truly met for many years.

By the 1930's the use of hard shell helmets in international and grand prix auto racing had become standard gear. The situation was similar in motorcycle racing. The very first of the modern hard "crash" helmet shells was not constructed of rigid leather but made up of several layers of cardboard held together with glue. Later, linen or other fine cloth impregnated with varnish resins was used. The sheets were laid up in a simple inverted dome-shaped mold and the "composite" was then allowed to cure to a solid shape. With a shape familiar to certain kitchenware, the name "pudding bowl" was often applied to these early helmets.

The first recorded use of the hard shell type of helmet in automobile racing in the United States was by Wilbur Shaw in 1932. Sir Malcolm Campbell had apparently given him an English made then-state-of-the-art helmet. The following year in a race at Legion Ascot Speedway in Los Angeles, California, the helmet was credited with saving Shaw's life during a crash. Within the next few years the use of helmets by US drivers became mandatory (Wagar et al, 1980).

Meanwhile, the football helmet industry was making strides of its own. John Riddell, a former high school mathematics teacher and football coach, had founded the Riddell Corporation in Chicago in 1929 to produce football related equipment. In 1939, Riddell introduced what appears to have been the first helmet to employ a molded plastic shell. Unfortunately, early production methods were not that good and cracking of the plastic shells during game play gave a bad reputation to plastic helmets for the first few years.

In 1941, Riddell patented an ingenious arrangement of fabric straps designed to keep the rigid plastic shell off the wearer's head but, far more importantly, provided a means for absorbing impacts to the shell. Sounding like a fundamental understanding of certain biomechanical principles, the patent itself states;

"....a shock at any point about the surface of the helmet is not transmitted directly to the wearer's head in the vicinity of the blow but is transmitted by slings, and thus spread over a large area of the wearer's head."

The hard shell Riddell suspension helmet debuted in 1949 in the NFL. Other sporting applications would, in time, pick up on this design concept. Many other football helmet manufacturers however took a different approach.

In a patent US 2,634,415(Turner and Harvey, 1953) the first padded hard shell football helmet was described. Claiming that *"..tape or strap suspensions have been sadly inadequate to the avoidance of severe head jolts."*, they proposed a resilient closed cell rubber-like foam be placed throughout the shell interior. Cavities in the liner aligned with holes in the shell were provided for ventilation. Encased in leather, this liner inside the hard molded plastic shell proved to be reasonably effective in dissipating impact energy and was the model for the current modern athletic helmet. However, the design was very hot and heavy and not well ventilated and as a result, the web suspension design continued to prevail for many many years – until more rigorous helmet performance specifications came along.

Meanwhile, Lombard and Roth working in the military aviation field, had also deduced that the Riddell-like suspension system, no matter how finely tuned, wasted space between the wearer's head and the inside of the helmet shell that could be better utilized for impact management. Filed in 1947 and issued in 1953, their patent (US 2,625,683), would alter crash helmet design in ways that have basically not since changed. Their idea was to fill, as completely as possible the gap between the head and the shell with crushable, energy absorbing material such as polyurethane foam. Clearly, though energy absorption was improved, this was not a repetitive impact kind of design as performance degraded significantly with subsequent impacts.

In order that their concept might achieve acceptance beyond the aviation community, Lombard and Roth formed Toptex Corporation in 1954. The plan was to "mass" produce motorcycle and auto racing helmets with crushable energy absorbing polyurethane liners. Importantly, at some point in time, a non-polyurethane liner material, expanded polystyrene bead foam, was selected for these helmets. This material EPSB foam was cheap, readily available, relatively easy to manufacture, light weight, its mechanical properties could be fine tuned but most importantly, it crushed more or less completely upon impact – and stayed crushed. The motorcycle, auto racing helmet as we know it today was born. To this day, virtually all helmets of the "vehicular" genre (i.e. crash helmets) employ this very same material.

During 1954, in what would become perhaps the preeminent helmet producer in the world, some auto racing enthusiasts began to manufacture helmets in a garage behind the Bell Auto Parts store in Southern California. The initial helmet shells were hand laminated fiberglass resin composites with a thick semi-rigid foam polyurethane liner. They were the first to extend the pudding bowl to cover more of the head in a 'jet" helmet style. Along with their extended coverage of the head, they were believed at the time to be among the most protective race helmets ever designed. Then along came George Snively.

Snively a physician and sports car racing enthusiast was present in 1956 at such a race when his friend William "Pete" Snell died from head injuries suffered in a racing accident. Subsequently he undertook a study that had never before been contemplated; a study that could only have been conducted at a medical facility. A biomechanics of head injury study that, though crude by today's standards, had a profound impact on modern helmet design and performance (Snively, 1957). In this article, Snively discussed his testing of helmets then currently available to the racing community. In what must be considered one of the most bizarre yet important experiments of its time, he had helmets placed on the head of cadavers and subjected each to a massive impact. Six cadaver/helmet experiments in all were conducted on six different brands of helmets then available. In every case but one, the helmets failed to prevent the cadaver head from sustaining what would, in a living human, be a life threatening skull fracture.

The only helmet to not result in a skull fracture was the helmet being made by Lombard and Roth's company Toptex. Though all helmets tested had a hard outer shell, this was the only one with a non-resilient EPSB foam liner and it was the one that would change the face of auto and motorcycle racing, cycling and equestrian helmet design, over the next 50 years.

Commensurate with these developments, Snively founded the Snell Memorial Foundation the sole purpose of which was to develop and promulgate performance standards for auto and motorcycle racing helmets. Today, the Snell label still appears on many high performance helmets in a variety of fields.

HELMET PERFORMANCE STANDARDS

A brief history of helmet standards development has been provided by Becker(1998) and will not be repeated here. However there are some highlights that must be acknowledged.

As noted earlier, performance standards for sporting helmets actually have their origins in the first of all such standards – that of the British Standards Institute BS 1869-1952 Crash Helmets for Racing Motorcyclists. This was followed in 1954 by BS 2826-1954, Protective Helmets and Peaks for Racing Car Drivers. Both standards provided tests to determine the energy dissipation capability of a helmet when subjected to a known impact. Standards for non-racing motorcyclists soon followed.

Biomechanically speaking not much was known or utilized in these early standards. However the importance of proper anatomical shape of the head was appreciated and a table documenting the polar coordinates of a test headform, for a medium sized head, was presented. These same dimensional characteristics are employed to this day though they have been extrapolated to a variety of sizes. Their origin however remains something of a mystery. Furthermore, given the shape of the "pudding bowl" helmet, there was no need to have such dimensional precision below the hat band region so only the top of the head was modeled.

From the point of view of test headform biofidelity, a very topical concern, nothing was implemented. In fact the original test headforms were constructed of wood – wood with rather precise characteristics vis *"...a wooden headform shall be made up of horizontal laminations of birch having a density of 40-45pounds per cubic foot at a moisture content of 12 percent. The wood shall be straight in grain, free from defects and free from dote."* Clearly this was no attempt to model the mechanical characteristics of most human heads. Rather it reflects the British engineers' fascination with properly stipulating a reproducible piece of test equipment. (This author has found no person that knows what actually is dote).

In terms of a failure criterion, today there is an appreciation that it should somehow reflect human tolerance values. In the 50s, it was known that when certain helmets were tested according to the BSI protocol – which involved dropping a wooden block onto the helmeted headform, some helmets transmitted higher forces than did others. On the assumption that lower would be better, the failure level was set at 5000pounds force. If one considers this in terms of inertia loading to a free headform, we're talking nearly 500Gs acceleration – a value that few would argue would not be fatal.

In the US, the first foray into the sports helmet standard issue was pioneered by Snively, Following up on his "skull busting" work, in 1959 he virtually single handedly published the first North American standard for racing crash helmets (Snell Memorial Foundation, 1959). His approach was to measure acceleration directly using a freely moving 12pound headform. Using the BSI standard as a guide, the failure criterion was set at 400G (equivalent to 4800pounds force) - again, probably a fatal level. The test headform that he employed was not wood but rather was described as being of K-1A magnesium alloy. Dimensional characteristics were not specified. Other than weighing 12 pounds, there were no other biofidelity requirements and there was no consideration regarding different sizes of heads and helmets.

Substantiating what would now be considered a high failure criterion, 400G, Snively with Chichester supplemented the data with a groundbreaking analysis of accident-involved helmets (Snively and Chichester, 1961). By replicating the damage to 10 helmets worn by racing drivers who had survived a crash, they deduced that; *"Survival limits of localized head acceleration of brief duration in man have been shown to exceed 450G."* Nearly 30 years later, Hopes and Chinn (1989) used a similar helmet damage replication technique and determined that *"Current helmets are too strong, and their design is optimized for an impact severity that gives little chance of survival."* That is, 400Gs is too high.

Typical of the pioneering efforts to relate head injury biomechanics to engineering considerations of helmet design and performance were those of Snively and Chichester (1962) and of Lombard and Advani(1966) some 40 years ago.

By 1961, the American Standards Association ASA had established a helmet standard committee and by 1966 had published its first standard for "Vehicular Users". (This standard was the forerunner to today's US DoT standard 218 for motorcyclists). The test procedures were essentially as Snively had first devised but some additional consideration of biomechanical tolerance levels was given. The WST tolerance curve had now been published (Gurdjian et al, 1964) and, notwithstanding other shortcomings, it pointed out that there should perhaps be time limits on acceleration exposure. The standard retained the 400G maximum limit but set time duration limits at lower levels. These so-called "dwell times", required that acceleration in excess of 200G not persist for longer than 2msec and at 150Gs for no longer than 4msec. The headform meanwhile remained the rather un-anthropomorphic K-1A alloy. In 1970, Snell upgraded its standard permitting only 300Gs maximum but with no dwell times. Which of these two failure criteria provided more protection continues to be debated today for some 35 years later, these exact same criteria are in place in modern crash helmet standards.

Football helmets took a different approach. Encouraged by the ongoing research at Wayne State University, athletic products manufacturers decided to try to implement some of the research findings to improve helmet performance. To this end, the National Operating Committee for Athletic Equipment was formed in 1969. NOCSAE designated Voigt Hodgson of the Gurdjian-Lissner Biomechanics Laboratory at Wayne state as its chief investigator and two significant departures from contemporary helmet testing occurred.

To begin with, Hodgson recognized the likely importance of headform biofidelity and with others (Hodgson et al, 1972) developed a test headform to better model the dynamic response of the human head. Based upon cadaver data the headform modeled both the geometric, inertial and frequency response characteristics of the head quite well. Originally in one size only, several sizes were subsequently developed. Like the human skull, and unlike any headforms before it, the NOCSAE headform could fracture if impacts were too severe – not a particularly good feature of a piece of test equipment.

The second important advancement was to employ the Severity Index (Gadd, 1966) as a measure of helmet performance. The failure criterion, in keeping with Gadd's view was initially set to SI=1000. This turned out to be too severe for helmets being produced at the time and the criterion value was moved upward to 1500. In 1973 NOCSAE published its Standard Method of Impact Test and Performance Requirements for Football Helmets. Since about 1980, virtually every football helmet sold and used in the United States has had to meet the NOCSAE standard. The failure criterion has since been lowered to SI=1200.

The introduction of the NOCSAE standard had a profound effect on head injuries in football in the US. Since its implementation, serious head injury rates dropped of the order of tenfold. Interestingly, the rate of concussions has not seemingly changed to such a great extent.

CURRENT DEVELOPMENTS AND THE FUTURE

Recent work with football players in the National Football League, has pegged the concussion threshold at a much lower value of SI than had previously been accepted (Newman et al, 2003). Furthermore, replication of actual incidents where players were concussed has led to the suggestion that the NOCSAE helmet test protocol could use updating. To this end NOCSAE has recently introduced improved headforms and working with the NFL is currently examining the manner by which its impact test methods and failure criteria might be improved (Withnall and Shewchenko, 2005).

In regard to motorcycle helmets, huge strides were made in Europe with the publication of the COST 327 report (2001). And the new ECE Reg 22 -05 motorcycle helmet standard now represents the state-of-the-art in performance specifications in this area; partly because of biomechanical considerations. Importantly, though from a test repeatability perspective, contentious, the test allows the headform to freely fall without any of the usual customary (guided free fall) constraints of contemporary protocols. This, it is argued, allows the head to respond in a more human-like fashion better permitting the implementation of human based criteria. Halldin et al (2001) have proposed an oblique impact test to address these same concerns. Unlike many other standards (NOCSAE excepted), each of the five ECE test headforms comprise the entire head rather than the partial headform employed by others. The anthropometrics are considered good and various sizes of appropriate mass and mass distribution are used. No attempt has been made to have biodynamic fidelity as this is now generally considered to be unimportant in helmeted impacts. This standard also sets the maximum allowable headform acceleration at 275Gs and (though the US Department of Transportation proposed and then revoked the idea of incorporating HIC in its performance standard in 1974) ECE Reg. 22-05 does in fact require that HIC not exceed 2,400.

A newly published standard for Formula 1 race car drivers (FIA 8860-2004), also employs maximum acceleration and HIC as failure criteria even though it employs the Snell-like guided free fall methodology.

One area that continues to frustrate the evolution of better helmet performance standards, and thus perhaps better helmets, is the rotation issue. Biomechanically, there is little argument today that Holbourn (1943) was right and it is rotational motion that dominates the nature and extent of brain injury NOT simply linear translational acceleration. Indeed, nearly 15 years ago, tolerance curves relating various types of brain injuries to rotational motion of the head have been proposed (Margulies and Thibault, 1991). In 1986, the author introduced a head injury assessment function that endeavored to take into account both translational and rotational acceleration (Newman 1986). This function, GAMBIT, has recently been employed in an effort to better understand the nature of protective headgear for soccer (Smith 2005). Several years ago, the author with colleagues Shewchenko and Welbourne (Newman, et al 2000) introduced a more general head injury assessment function, the Head Impact Power HIP. This function, which considers the maximum rate of translational and rotational energy transfer to be the controlling element in inertially induced brain injury, has been successfully used in the development of a new North American football helmet (The Riddell Revolution). HIP is also currently being employed to help quantify head injury threats in soccer (Shewchenko, et al, 2005).

Still, the failure criteria for every published helmet performance standard in the world continue to be based solely on linear acceleration of a test headform. How we will reconcile this conundrum remains to be seen. One thing is certain, helmets are better than they were 50 years ago and this is partially due to the recognition of certain basic biomechanical concepts. However, little improvement can be expected for the next 50 years until we determine ways to implement more of what little we know about the mechanisms of head injury.

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