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Measuring *in situ* Head Accelerations and Evaluating Clinical Outcomes in Collegiate Football Players

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ABSTRACT

Mild traumatic brain injuries have serious immediate and long-term effects on sports players. The purpose of this study is to measure the accelerations of the head during football impacts in order to reduce the risks of brain injuries in the future. Football was chosen because it is a head impact rich environment. A newly developed in-helmet six accelerometer system that transmits data via radio frequency to a receiver with a laptop unit on the sideline was implemented. The system can instantaneously provide the researcher with a visual representation of the immediate and cumulative game's impacts of up to 64 players. From the data transfer of these accelerometer traces, the sideline staff has the head acceleration resultant, the head injury criteria value, the severity index value, and the direction and magnitude vector of the impact location. One game's results are described in this paper to show a sample of the information being collected in this study. For eight players there were 347 total impacts with an average of 21.5 ± 19.7 g, 11.5 ± 33.4 HIC, 16.7 ± 51.2 GSI, 769.9 ± 1082.7 rad/s^2 rotation about the x-axis and 1382.8 ± 1547.3 rad/s^2 rotation about the y-axis. This in-helmet accelerometer system proves to be very useful in collecting accurate data that quantifies head accelerations for injured and non-injured players.

INTRODUCTION

Traumatic brain injuries occur in 1.5 million people in the United States each year, of which 75 percent are mild traumatic brain injuries (MTBIs) like concussions (Gerbeding, 2003). Sports related concussions constitute 300,000 of these injuries (McCrea, 2003). In the 2002-2003 college football season, 8 percent of all injuries were concussions (Guskiewiez, 2003). Football has the most total concussions of any sport and has had an increasing rate of injury in the last seven years.

Research on concussion threshold values for football players has been advancing for years in order to give more insight into the cause and effect of a mild traumatic brain injury. In the early 1970s, Moon and Reid (1971 & 1974) instrumented the headbands of suspension-style football helmets with a frequency modulation (FM) based accelerometer and electroencephalogram (EEG) system. Morrison (1983) utilized a similar system in the early 1980s at Penn State, although without the EEG capabilities. While laying the ground for future research and providing a proof of concept, these studies were limited in that the accelerations measured were often of the helmet shell as opposed to that of the head. In 2000, Naunheim (2000) instrumented hockey and football helmets with a padding-embedded tri-axial accelerometer. Though Naunheim's (2000) results were more realistic than Moon and Reid's, they were limited from an inability to ensure that the accelerations were accurate for the head.

Recently, methods of head acceleration measurement have utilized a combination of video analysis and dummy reenactments of impacts from game film. Newman (1999 & 2000b) and Pellman (2003) published a series of papers based on a National Football League (NFL) study of MTBIs in professional football. In this series they studied concussive impacts that were recorded on film from two or more different angles. They used this video data to reconstruct the angle of the impact, speed of the impact, and the resultant player kinematics. This data provided the information necessary to recreate the impact with instrumented Hybrid-III dummies. While this study established linear risk curves, rotational risk curves, and impact tolerances for concussion, the methods used are limited and time-intensive.

The dependence on video reconstruction and dummy reenactments not only introduces error, but it prevents the data from being used clinically on the field. The previous research has not given accurate real-time information on the direction and magnitude of the impacts football players receive. The purpose of this paper is to introduce a new method for accurately collecting and analyzing real-time head impact data in collegiate football.

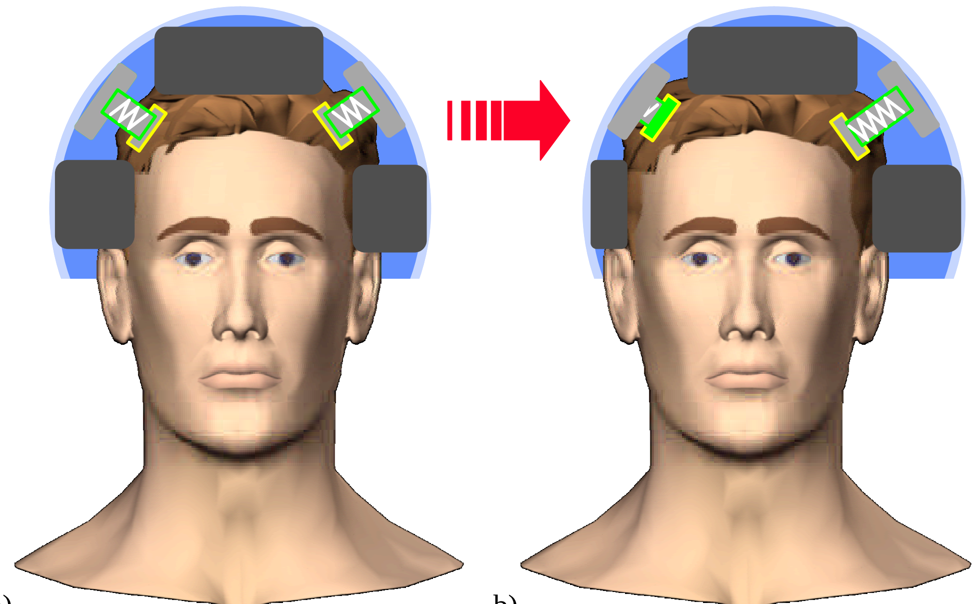
METHODOLOGY

This study utilizes the Head Impact Telemetry System (HITS) (Simbex, Lebanon, NH), a wireless system that provides real time data of impacts to a signal receiver located on the sidelines. Each monitored player wears an in-helmet unit (Figure 1) designed to be fitted into a large or extra-large Riddell VLR-4 (Figure 2) football helmet. Spring mounted accelerometers keep constant contact with the head to ensure measurements are of head rather than helmet accelerations (Figure 3). The in-helmet unit has six single axis accelerometers operating at 1000 Hz sampling frequency, a circuit board that retains the last 32 impact data sets, a connection for a battery pack, and an antenna for transmitting the data.

Figure 1: Six accelerometers, FM-antenna, and rechargeable battery pack.

Figure 2: Virginia Tech helmet with the accelerometer system installed.

If any one accelerometer or the resultant of all six goes over 15 g, the data is saved for 12 ms before and 28 ms afterwards. The resultant of all six accelerometer traces is a vector representing the actual head acceleration for 40 ms in units of gravitational acceleration (g). The location of the hit and two of the three rotational accelerations are also computed from the accelerometers. HITS was validated at Wayne State University using a series of impact tests with a helmet-equipped Hybrid-III dummy that featured a 3-2-2-2 head accelerometer array. HITS correlated well with the 3-2-2-2 data (R^2=0.97) and had a ±4% error for linear and rotational accelerations as well as HIC scores.



a) b)

Figure 3: a) The acceleration sensors maintain contact with the head in a properly worn helmet. b) The sensors measure head accelerations as opposed to helmet accelerations.

In addition to the in-helmet units, there is a sideline unit that downloads the data in real time. However, because each helmet can store up to 32 impacts between transfers with the sideline receiver, constant communication is not necessary. In this study, the receiver was located at the 20 yard line of the visitor's sideline and had an approximately 70 yard range. The six single accelerometer traces and analysis are presented to the sideline unit controller on the laptop screen in four frames: the vector showing the current impact direction, the game's total impact vectors, acceleration versus time graph of traces, and an acceleration magnitude bar chart (Figure 4).

Recorded data is time-stamped and analyzed post game. The data acquisition time stamp is correlated with video footage taken during both practices and games. The practices allow additional impact events to be collected. During any practice or game up to eight players were tracked simultaneously. However, there is capability for tracking 64 players at a time. Players are selected each week by the medical staff to provide a representative cross-section of all football players. All players selected have signed an Institutional Review Board form for Virginia Tech and Edward Via Virginia College of Osteopathic Medicine.

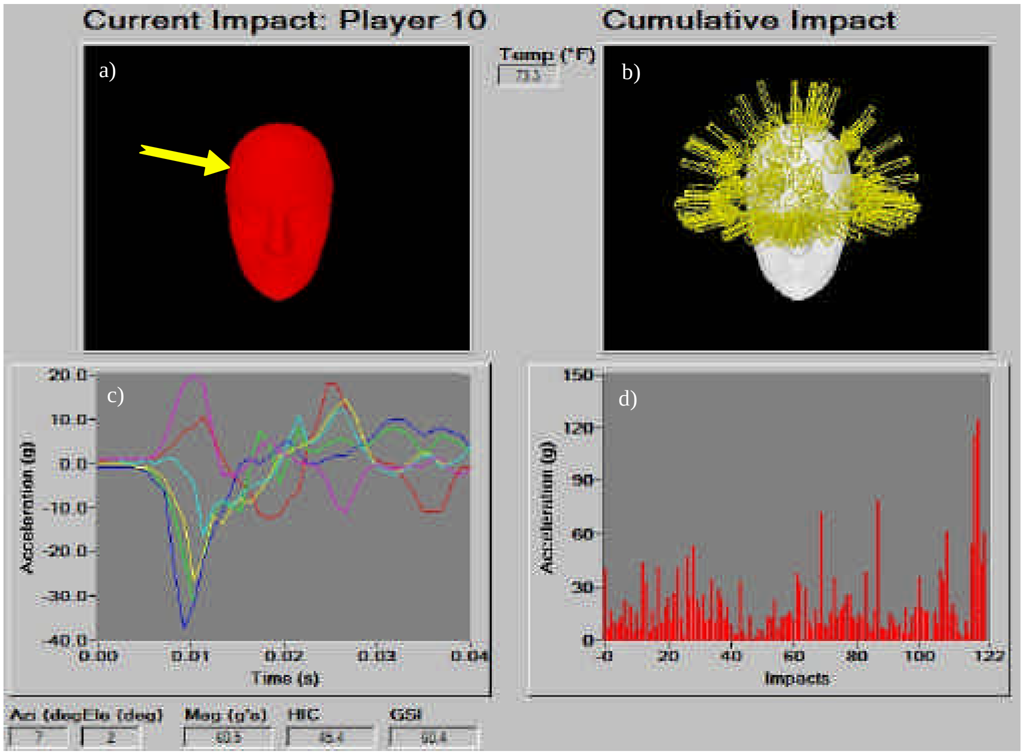


Figure 4: Sideline controller laptop screen. a) A directional vector indicates the most recent hit location. b) Cumulative vectors for that day are shown on the model head. c) Six accelerometer traces are plotted versus time for 40ms. d) The bar graph shows the acceleration resultant value in the direction of the hit.

The sports medicine staff maintains detailed player histories, including pre-season neuropsychological records. HeadMinder™, a web based assessment program, is used pre-season and post injury to evaluate a player's cognitive status. Physical indicators, such as postural sway, confusion, or headache are closely observed and measured by the doctors. When an injury is more serious or persistent, there are more elaborate neuropsychological tests and neuroimaging procedures available.

RESULTS

The results presented are for eight players during one game. Results are from a defensive back, lineman, running back, and wide receiver. The average for all eight players is 21.5±19.7 g, 11.5 ± 33.4 HIC, 16.7 ± 51.2 GSI, 769.9 ± 1082.7 rad/s^2 rotation about the x-axis and 1382.8 ± 1547.3 rad/s^2 rotation about the y-axis. The largest impact a monitored player received was 123.9 g, 306.7 HIC, 608.3 GSI, 9417.6 rad/s^2 rotation about the x-axis and 11493.0 rad/s^2 rotation about the y-axis. No head injuries occurred in this game.

Of the eight players monitored, the two linebackers received the largest amount of hits, between 60 and 70 each. The defensive back, offensive lineman, running back, and wide receiver each had less than 40 impacts above 15 g. The difference in impact magnitude is illustrated between the fullback and defensive lineman (Figure 5 and Figure 6). The fullback in this analysis saw three impacts over 80 g while most other impacts stayed under 20 g. The fullback endured a maximum acceleration of 111.9 g. The defensive lineman had many more impacts in the range of 15 g to 40 g. However, this player had only four impacts over 55 g with a maximum impact of 65.7 g.

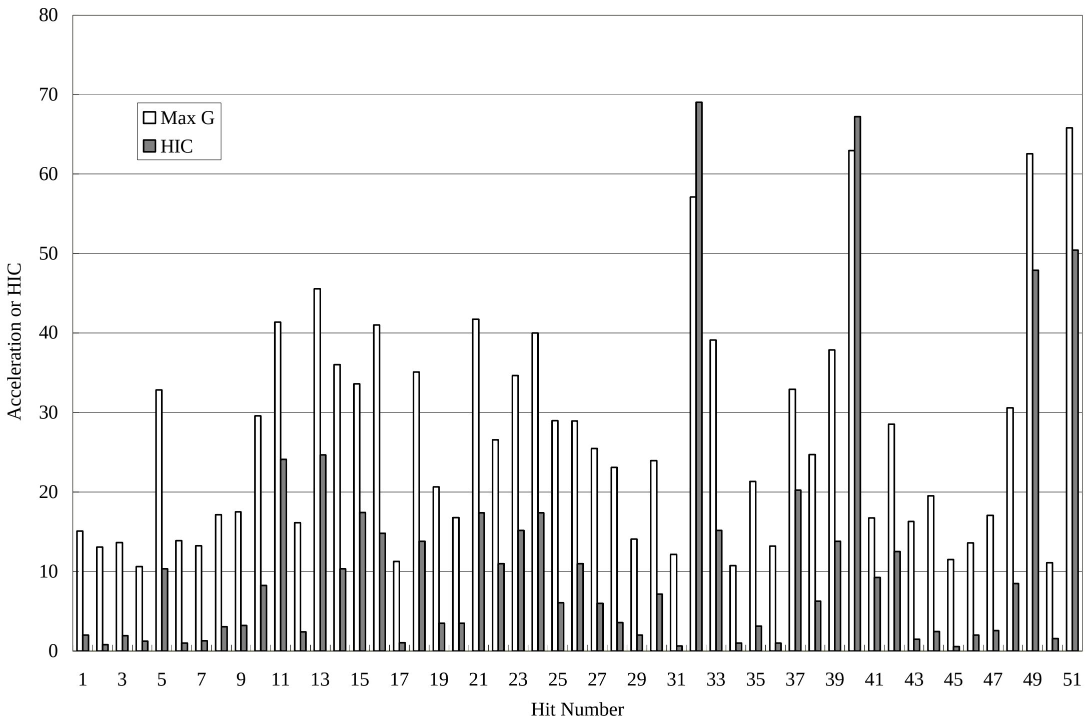


Figure 5: All max acceleration and HIC values for one defensive lineman in one game. Hit number indicates the order in which the impacts occurred.

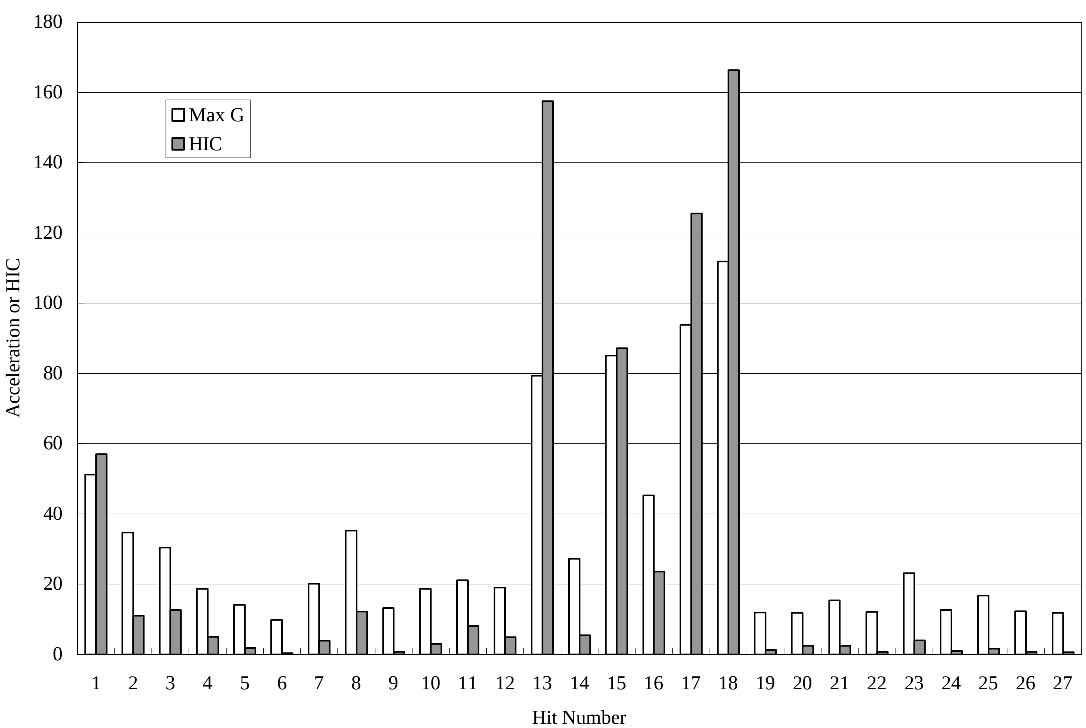
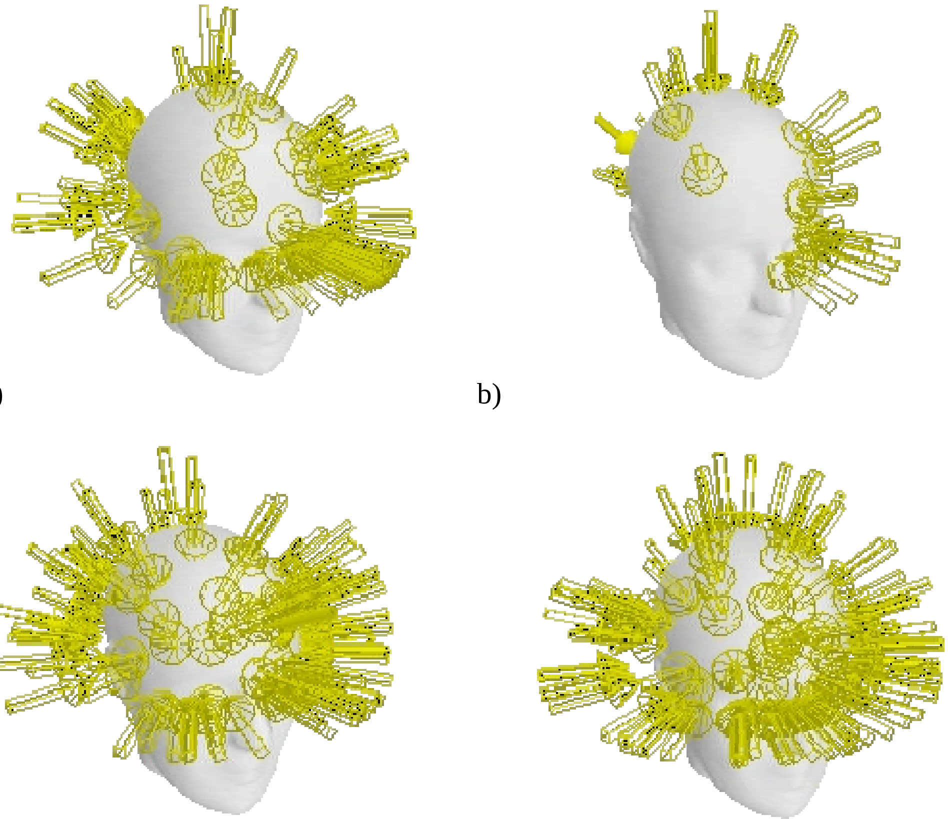


Figure 6: All max acceleration and HIC values for one fullback in one game. Hit number indicates the order in which the impacts occurred.

The locations of all impact vectors are shown on a model head for four player positions (Figure 7). The receiver exhibited a Mohawk pattern of impacts with very few lateral impacts. This pattern is seen in defensive backs as well. Conversely, the fullback exhibits a lateral band of impacts circling the equator of the head, with a second concentration of impacts at the forehead. In addition to impacts similar to a fullback, the linebacker and defensive lineman have more impacts to the rear and upper halves of the helmet.



a) b) c) d)

Figure 7: One game's cumulative impact vectors for four positions. a) Fullback, b) Receiver, c) Defensive Lineman, d) Linebacker

DISCUSSION

Pellman (2003) states that the peak acceleration for a concussive impact is 98 ± 28 g with a 15 ms duration. The acceleration magnitudes from the HIT system agree with this finding of Newman (2000a & 2000b) and Pellman (2003) more than with the Reid (1971 & 1974) and Moon numbers, despite the testing similarities between the HIT system and Moon and Reid. However, the HIT study has an advantage over Pellman (2003) and Newman (1999 & 2000b) in measuring many non-concussive impacts which have featured acceleration magnitudes and injury criteria scores higher than 98 ± 28 g.

The location of impacts is also in agreement with Pellman (2003). The majority of the impacts for the hitting player are located at the face or forehead, while the primary locations for the player being hit are to the side or back of the head. A new discovery from this study is that the location of impacts is also position-dependent, with receivers exhibiting an entirely different location profile than linemen.

The primary finding of this study is that the number and magnitude of head impacts are much higher than previously anticipated. A number of players exhibited one or more impacts greater than 90g in a game or practice. The medical staff noted that the players were aware that these impacts were larger than the normal impacts, but neither the players nor the staff was previously aware of the actual magnitude.

The primary limitation of this study is the difficulty of recording the most minor concussions. Unlike Pellman (2003), this study cannot *a priori* select for concussive impacts. An additional difficulty is that the data from the other player involved in a concussive hit cannot be tracked at this time, should the impact occur in a game or with a non-instrumented player. It is believed that with greater player coverage, and the expansion of this system to more teams, these obstacles will be resolved in the future.

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DISCUSSION

PAPER: **Measuring *in situ* Head Accelerations and Evaluating Clinical Outcomes in Collegiate Football Players**

PRESENTER: ***Stefan Duma, Virginia Tech - Wake Forest Center for Injury Biomechanics***

QUESTION: *Dave Meaney, University of Pennsylvania*
Stefan, I think it's a very interesting paper and it really adds, and I'm sure in the future will add even more to this volunteer database that we're really in need of, especially with respect to brain injury data. Had a couple of technical questions - you went quickly through it. You mentioned that you're getting two of the three rotational accelerations? Which planes?

ANSWER: We're getting x and y; we are not getting z. If you look at how the six are aligned, we do not get z. And like I said, this is the first season this has been implemented so we're looking at maybe adding different accelerometers or optimizing some of the algorithms to get that third one.

Q: And, you had mentioned you were + or - 4% with respect to the dummy data that you had from the Wayne group.

A: Right.

Q: Does that error change with magnitude? Does it begin to get a little bit larger at higher levels of--?

A: No, we do not see that trend. We see it more by direction, and the error is more direction and it has to do with the rotational. But when you get more of a z-axis rotation, the error starts to really get higher. But, the peak G, the HIC, and the GSI pretty much track 2 to 3 to 4% for all the different impact locations. All the data is available. And again, we're putting it together for these papers right now.

Q: Okay.

Q: *Guy Nusholtz*, *Daimler/Chrysler*
My experience with locating accelerometers on heads indicated to keep them from moving around, you had to screw bolts into the head, which I'm sure football players aren't going to engage in. I'm not sure that you're really--There should be a response characteristic of your spring in mass, which will move that thing around and it should slide to some degree during the impact. So, it's not really clear. Even with your claim of 4%, which your data really didn't look that the way you presented it, that there's probably a lot of error terms in this thing, which are confounding your results. Do you have any way to try and correct that either using official data or some other scheme?

A: Well, I mean on the bolt issue: I think after the West Virginia game, everything's fair game! So, we might go back to that.

Q: Then, that's a solution.

A: That's a solution. That's the kind of stuff we're looking at right now. Again, this is the first full season of it. If you look at the dummy tests, and the big problem, even with the NFL test, is that the dummy is not simulating these boundary conditions. So, how it slides on a player's head is very different than how it's going to slide on a dummy, even though they make some adjustments for that. I do not have a solid answer for your question other than it's what we're looking at right now with some of these algorithms.

Q: Okay. Thank you.

Q: *John Melvin, Tandelta Inc.*
Along the lines of Guy's question, I would imagine that the scalp, so-called, of the Hybrid III dummy is very well coupled compared to what real people's scalps are like. So pressing it against the scalp, you could still have some interesting problems. You might want to try those experiments with a little more realistic gel-like surface just to see. Another possibility--and I don't know about your SIMBEX system there and how it broadcast, but is it possible to put at least one accelerometer in their mouthpiece? They do wear mouthpieces and that's the best coupling you can ever get. And if your system doesn't predict what that thing says, then you have a problem. It would be a good thing to try and implement if you could.

A: Well, there's a couple answers. There's a lot we can do in a laboratory with verifying some of these systems, and that's what we're doing over the course of the next year. The mouthpiece is a problem and unfortunately, they don't always wear their mouthpieces. It's a much lower usage rate than you would expect. They forget. They lose 'em. They're in and out between games. And then, there's issues with the wiring, the battery, the power supply and things like that, so we won't really do that. What's more likely to increase some of this a different accelerometer at a different location. But, all of these things are what we're looking at now.

The other thing I wanted to say on the sliding issue: In the NFL data, they put a surface in between the dummy and the helmet to try to simulate some of that. The way our players are fit, and most of them are: They put kind of a soapy solution on their head. They put the helmet on. They blow up it and they grab the facemask. And until, until--with this solution, until the forehead skin moves with that facemask, that's as tight as they go. So, they're very tight. Does it eliminate the slipping? No. There's still some slipping, but it seems to be--especially seems to be within the range of error of what we can do from reconstruction.

Q: The other is: Like Dave says, it may be at the really heavy impacts is where they're gonna slip and give you the erroneous data. In the Indy racing league, they're running ear accelerometers in their drivers, and they have gotten an occasional data set. One driver in particular, in a rear impact, pulled 200 g's on his ear accelerometer with no concussion. We've modeled that kind of behavior down at Wayne State, and we estimate the HICs in that situation at about 2,000 without head injury. But with angular accelerations, it seemed to be the factor into whether the driver's injured or not. So, these are interesting. Everybody's looking at this. It'll be interesting to compare the racecar data with the football data. Maybe you should try an earplug accelerometer. I know football players--at least pros--wear earplugs now to listen to their coach, so that maybe a solution. I don't think they do that in collegiate, but it could be an earplug without a radio in it to try and--cause you can couple fairly well in the ear canal if you do it right.

A: Yeah, that's a good idea, but there's a problem in that there's so much communication on both sides of the line that they don't like to wear earplugs. They need to be able to talk. So, that's kind of the limitation of the earplug.

Q: Thank you.

Q: *Erik Takhounts, NHTSA*

Okay. A quick warning: Stefan, with machined nine-accelerometers into the dummy's head, they're very well attached and glued and everything. You take the same dummy. You run it once, you run it twice and it's consistent so it returns good kinematics. You run it third time, the head starts rotating like a badly thrown football. So, there's some small errors in your accelerations if you try to predict a 3-D rotational motion of the head. That's where your problem comes in. It's a 3-D motion of the head that's, that amplifies your problem tremendously. Maybe you should look at something simpler, something like maybe assume some kind of 2-D motion and all you need is just two accelerometers.

A: Well, I wouldn't want to sell ourselves short from the start. I mean, if you look at validating these tests, particularly on the CG acceleration, the direction and the magnitude correlates extremely well with what we see in the dummy test. Now, the rotation is certainly not there, perfected yet. I don't know if I'd want to step back. I mean, I think that there's been a lot of validation tests done with different head forms, different dummies, looking at the non-accelerometer rate and it seems to track very well. The 4% level, again, I think this has to be put into the context of all the work that was done with the NFL data is 15% error-bound. If we're within 4%, we're certainly lower accuracy than what they were seeing in some of that more elaborate style testing. So, I don't know if I would say I would necessarily agree that we should step back and look at two. I mean, I think the data seems to correlate very well with what the dummy predicts.

Q: I'm not sure you answered my question, but thank you.

A: Okay. I'm sorry. I'll talk to you later.