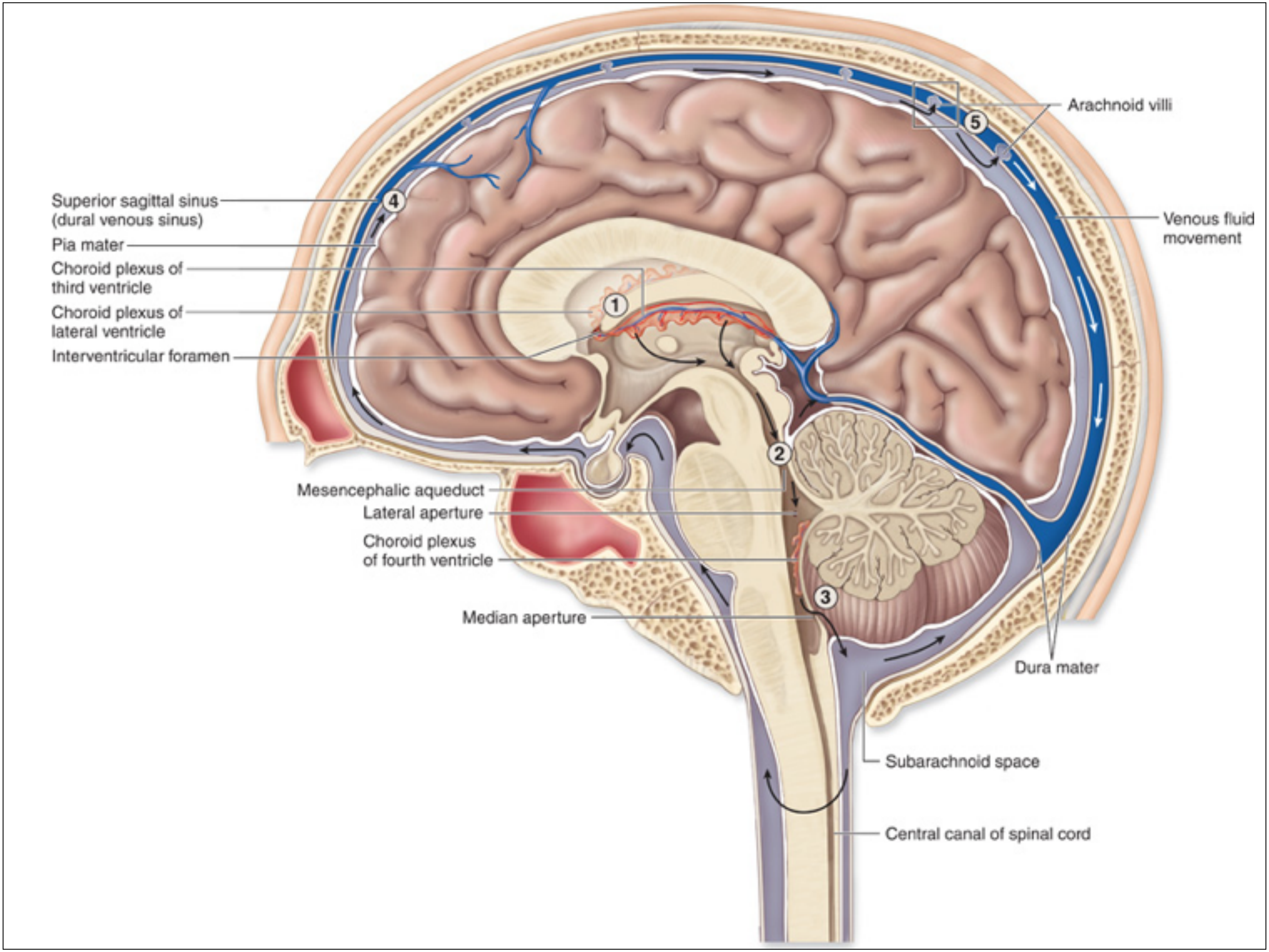
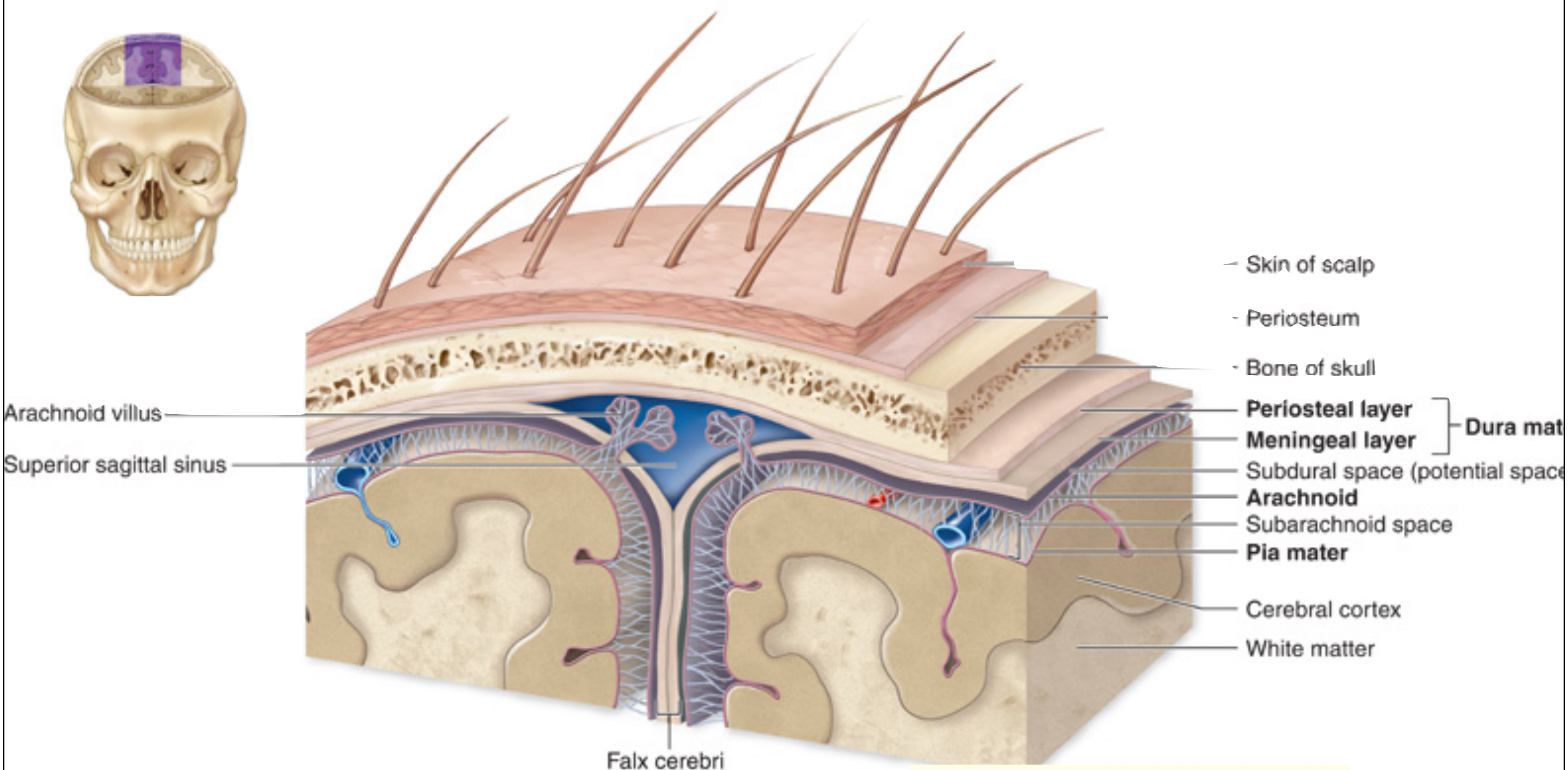
Closed TBI and its Protection: A Physics Perspective

Eric Blackman (U. Rochester)

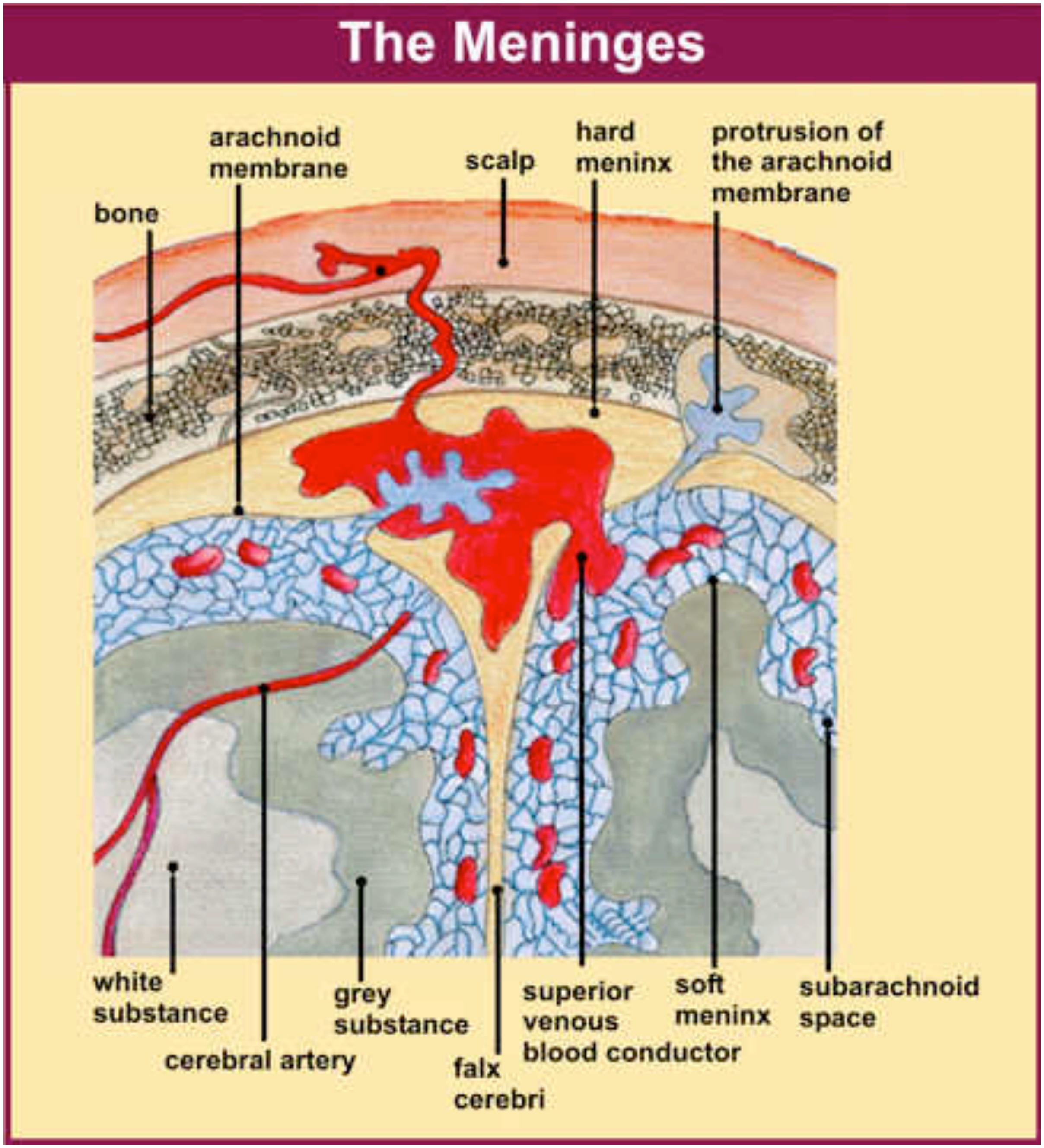
- Injury to the brain without skull fracture

- Contexts
  - head impacts
  - blast overpressure


Superior sagittal sinus (dural venous sinus)
Pia mater
Choroid plexus of third ventricle
Choroid plexus of lateral ventricle
Interventricular foramen
Mesencephalic aqueduct
Lateral aperture
Choroid plexus of fourth ventricle
Median aperture
Arachnoid villi
Venous fluid movement
Dura mater
Subarachnoid space
Central canal of spinal cord
1
2
3
4
5


Skin of scalp
Periosteum
Bone of skull
Arachnoid villus
Periosteal layer
Meningeal layer
Dura mat
Superior sagittal sinus
Subdural space (potential spac
Arachnoid
Subarachnoid space
Pia mater
Cerebral cortex
White matter
Falx cerebri

rrr


The Meninges
arachnoid membrane
scalp
hard meninx
protrusion of the arachnoid membrane
bone
white substance
cerebral artery
grey substance
falx cerebri
superior venous blood conductor
soft meninx
subarachnoid space

Closed Traumatic Brain Injury (TBI)

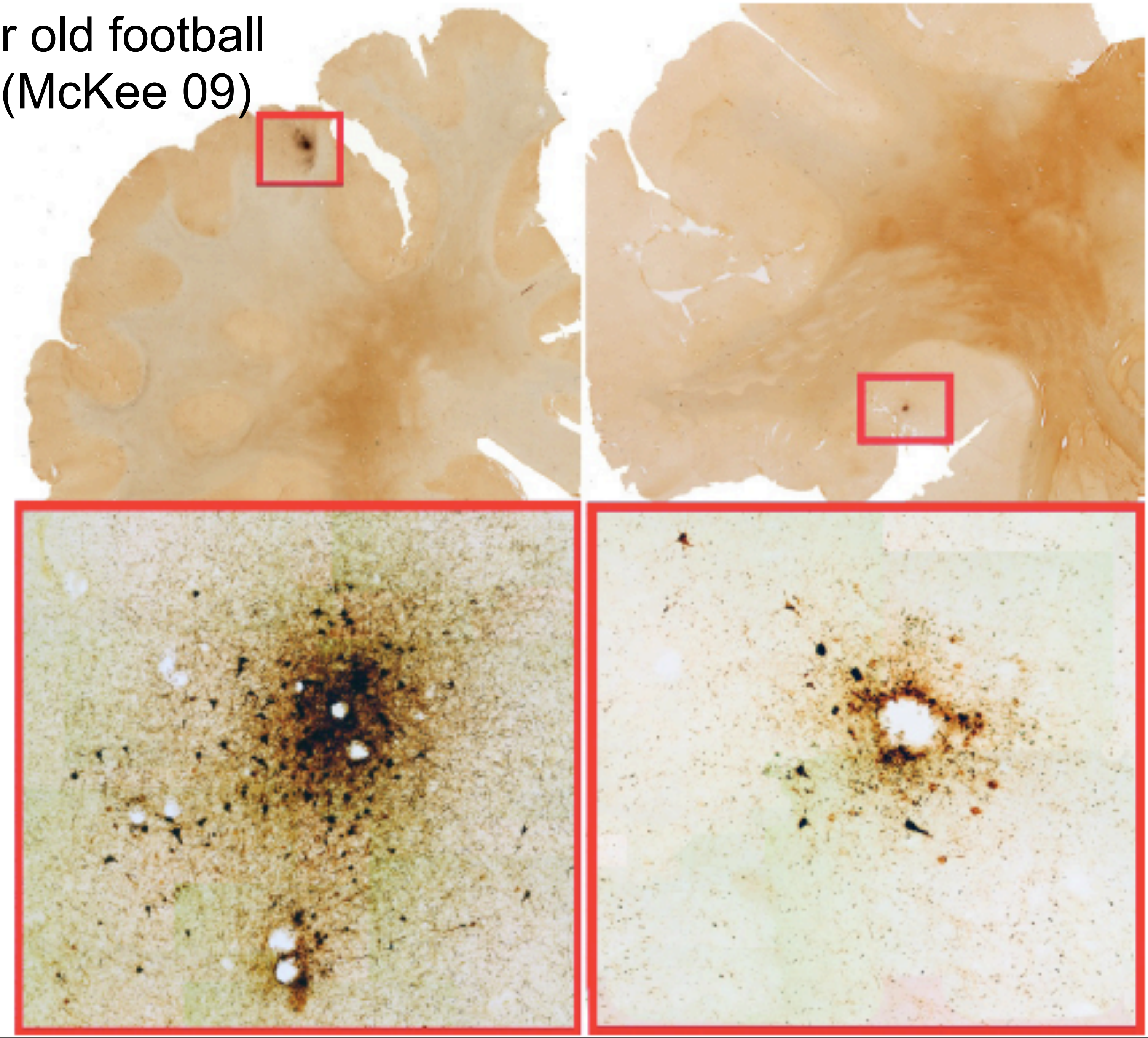
- **concussions** (non-local; midbrain, brainstem, frontal lobe)
- **diffuse axonal injury** (shear damage of axons; white matter grey matter linkage)
- **contusions** (general bruising)
- **subdural hematoma** (bridging vein damage)
- **chronic traumatic encephalopathy** (CTE) (degenerative brain injury from repetitive head trauma)

Chronic Traumatic Encephalopathy

(e.g. McKee et al. 2009; 2013..)

- CTE: toxic "tau protein" builds up in brain cells, preventing normal connections to other cells; cells die
- tau protein shows up as neurofibrillary tangles (NFTs) and glial tangles
- Tangles are formed by hyper-phosphorylation of tau proteins in microtubules, causing tau to aggregate
- accompanies dementia though not itself a signature of Alzheimers (no beta amyloid)
- Prevalent in brain tissue of deceased football players and boxers, and, soldiers (Goldsein 2012), a soccer player, baseball player, rugby player, wrestler (McKee et al 2013; Branch 2014); many without history of severe concussions.
- Role of repeated low level impacts is most serious TBI issue and least understood

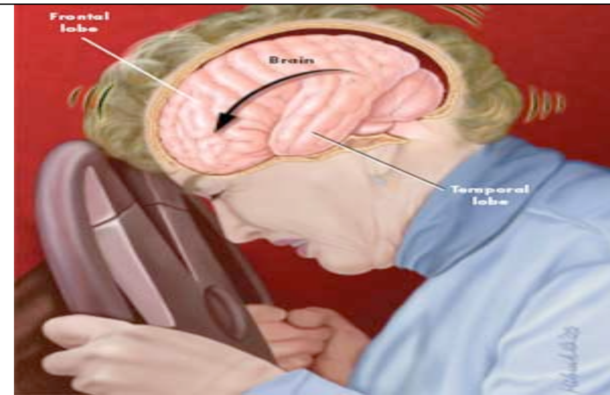
Brain of deceased 18 year old football player (McKee 09)



Costs of TBI

- **Human costs**
- **Civilian**: > 3.8×10^6 cases/yr; 50% auto; 25% sports (McArthur 04; Langolis et al 2006)
  - 20 deaths per 100,000: $20 billion/yr treatment
- **Military**:
  - before 2006; estimated 3% of soldiers have TBI (60% of hospital injured soldiers)
  - 0.6% of all soldiers serious TBI
  - New screenings: 2006-2009 ~20% of all troops have TBI; 1.5% of all troops unfit to return.
  - $2.7 million (Blimes 07) per 25 yr post-TBI lifetime; >$2 billion/year for treatment
- **Workforce / mission / security costs**

Head Impacts

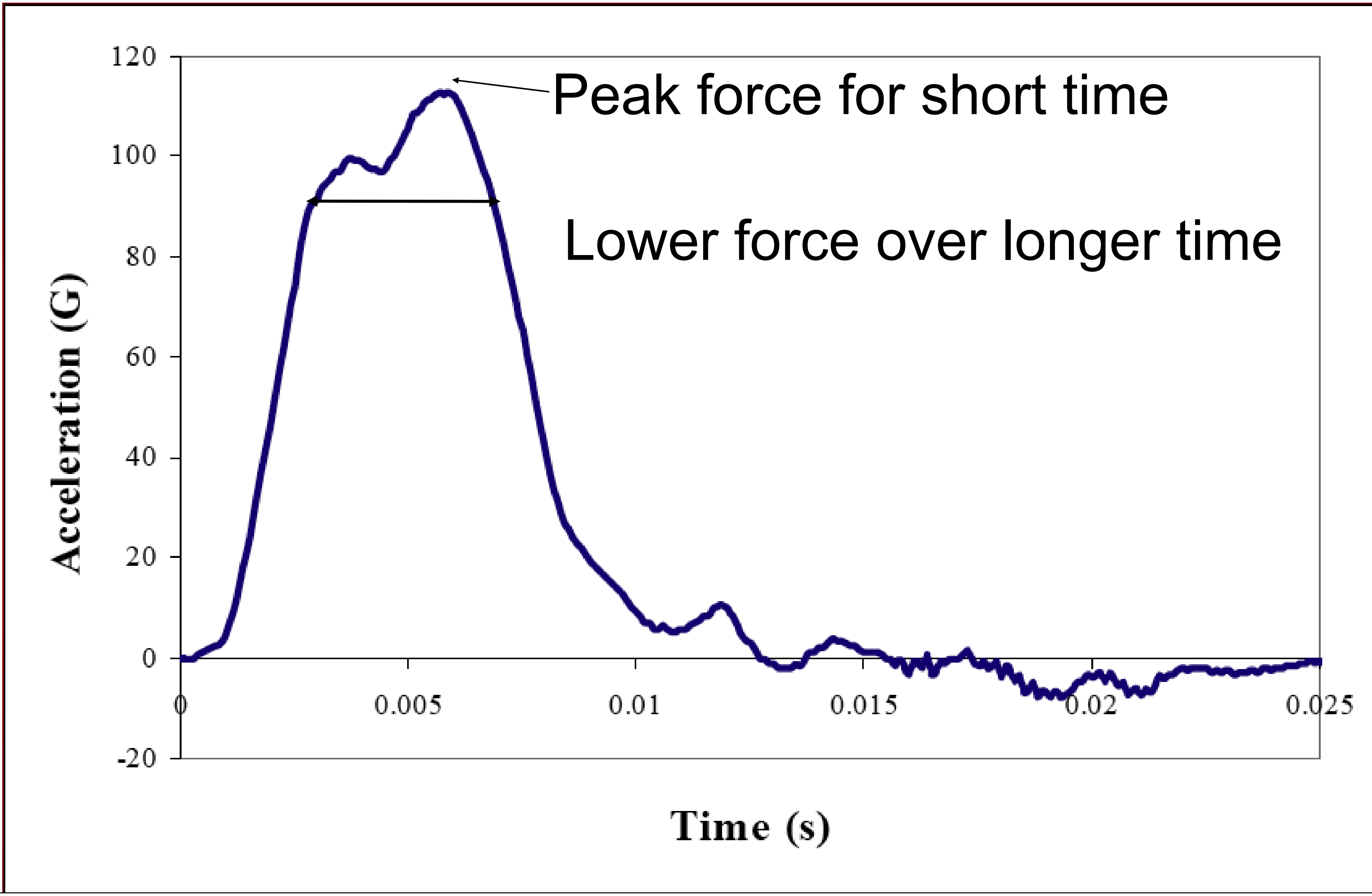


- Gravity or explosion converts gravitational potential energy or chemical energy into bulk kinetic energy
- Rapid deceleration of head on impact implies large force
- As head impacts, brain keeps moving
- Brain ‘crashes’ into skull displacing cerebral spinal fluid; stresses brain tissue both by compression and shear
- During impact, kinetic energy is converted into brain deformation energy
  - Brain damage arises because the kinetic energy is dissipated in brain rather than in helmet or skull
  - tissue stress / mechanical thresholds for injury
  - magnitude of forces vs. duration of forces
  - linear force, rotational torques, etc. what is the best metric? (not yet clear)

Role of Helmets for Impact TBI

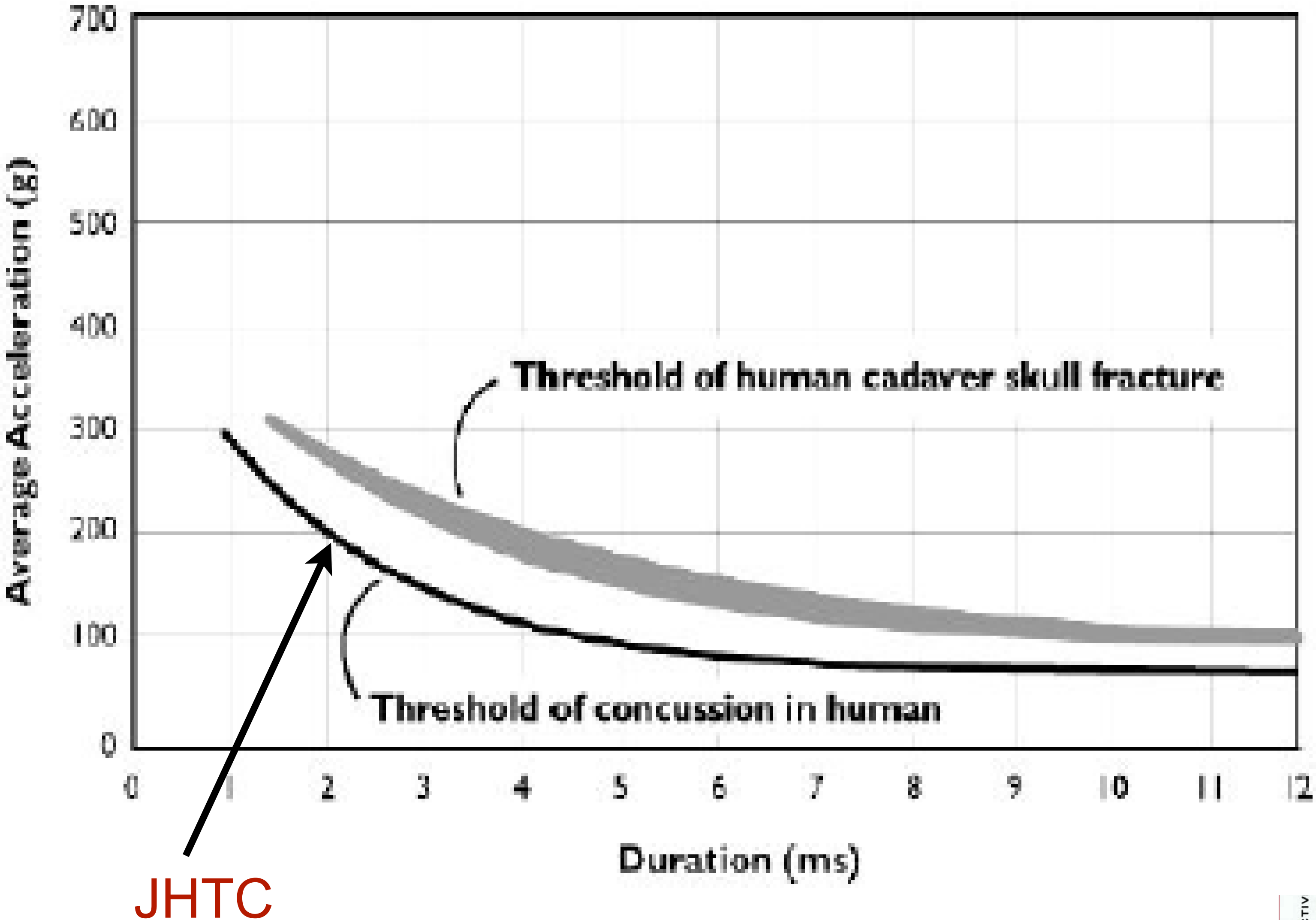
- Protecting skull from fracture is insufficient to protect brain from crashing into skull and distorting therein
- Hard shell alone is ineffective
- Need to:
  - reduce head acceleration (reduces force incurred)
  - reduce energy absorbed by brain (reduces energy available to sustain a distorted brain for extended period)
- Need cushioning to reduce head impact acceleration and thus force on brain
- Need also “somethng” to stop head rotation

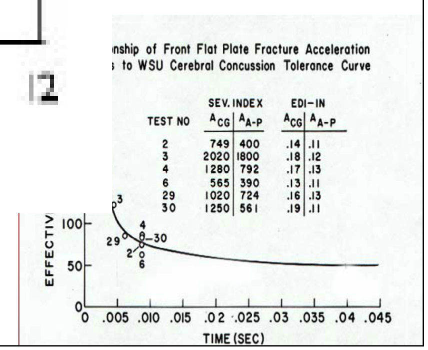
Impact Linear Acceleration Profile



Typical TBI/Blunt Impact Standards

Ono et al. 1980 (human cadaver and scaled monkey data)





Common Empircal Injury Metrics

1. "Peak g"= maximum linear acceleration

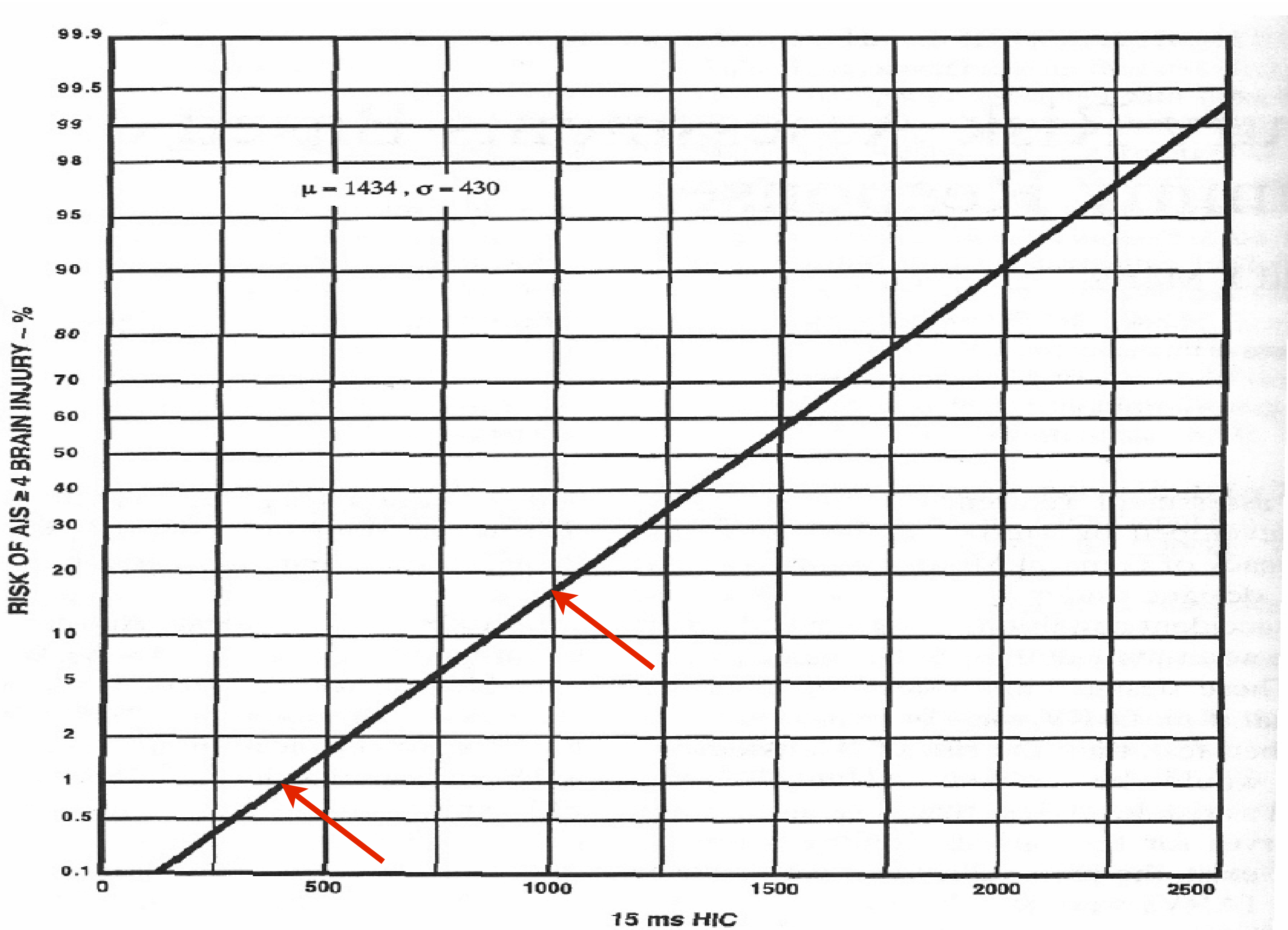
2. $$\mathrm{HIC} = (t_2 - t_1)\left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} a(t)dt\right]^{5/2} \quad \text{(around peak)}$$

3. $$SI = \int_0^T a^{5/2}\, dt \qquad (0 < \mathrm{t}_1 < t_2 < T)$$

4. Rotational acceleration (radians /sec^2)

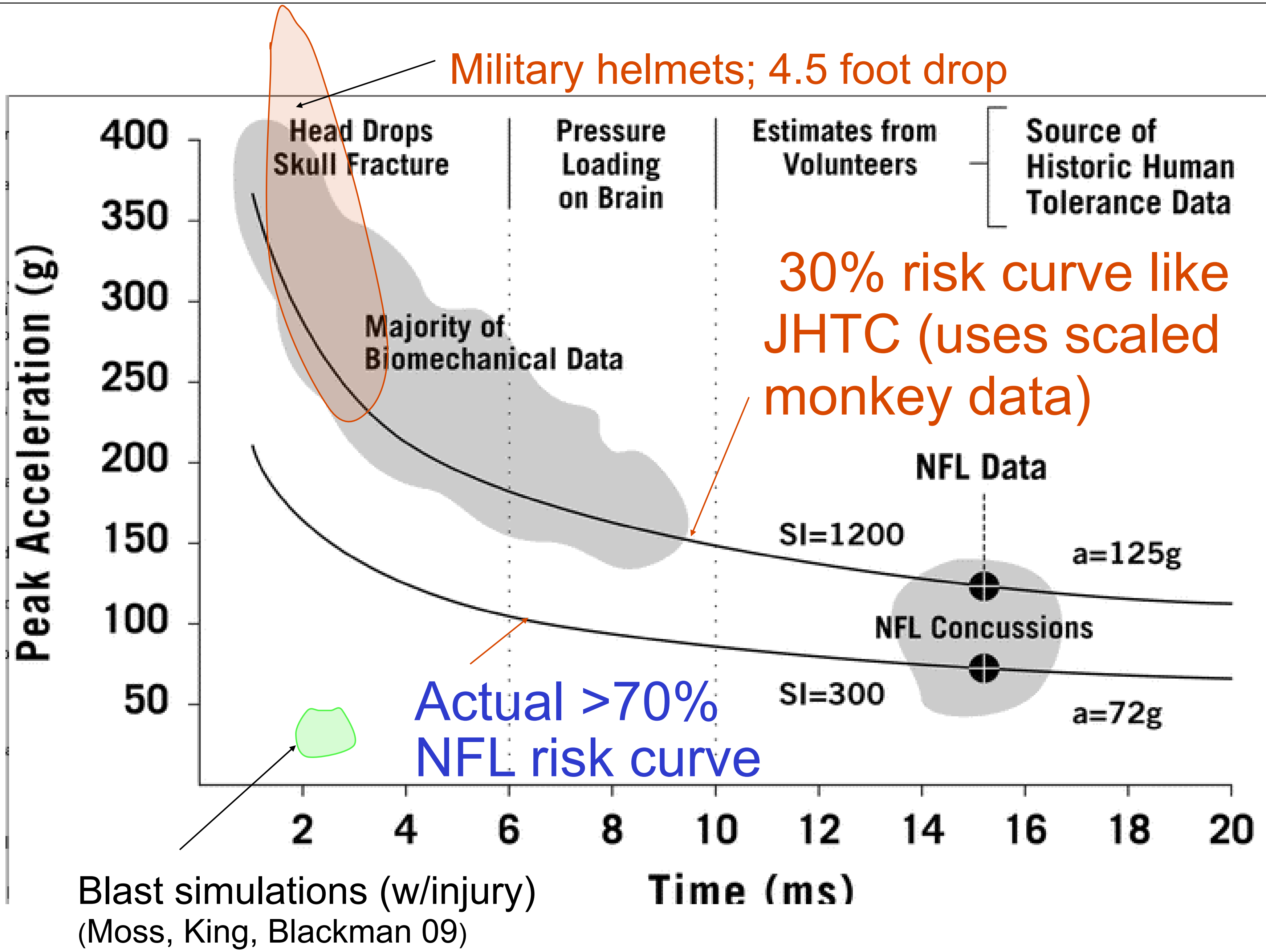
- SI (severity index) and HIC (head injury criterion) empirically accommodate acceleration and duration from cadaver and animal injury data
- Can create injury probability graph
- e.g.: Head HIC > 1000 (sec), 17% risk of potential life threatening TBI (and 1.7% actual risk of death, Prasad & Mertz 1985)
- HIC & SI are not derived from "first principles": a serious deficiency in the science of head injury protection
- No rotational metric is used in practice

HIC15 AIS4 Injury Risk (Prasad & Mertz 85, data compilation)



Helmet Standards for TBI are Lousy

- NHTSA uses HIC= 1000. (supposedly 1.6% chance of fatality 30MPH collision for restrained driver. Not a TBI standard.
- NOCSAE uses SI=1200; (~ JHTC) but for NFL does not protect against TBI: should actually be 140 based on concussion data
- standard for most military Helmets has been peak g standard and its lousy for impacts (and even less helpful for blasts)
  - Slobodnik (1980): need <150G at 1.5 meters drop
  - special forces helmets: standard 150G at 1.5 feet drop (2005)
  - Free falls of 3 feet for a ~5kg head form including PAGST or ACH helmets give 300G (McEntire et al.05)
- **NO standard for CTE**
- **NO serious standards for youth helmets; often helmets are re-used**
- **NO rotational acceleration standard**
  -



augmented from Pellman et al (03,06)

Simulations of Helmet Pad Efficacy

Moss, King, Blackman (2012)

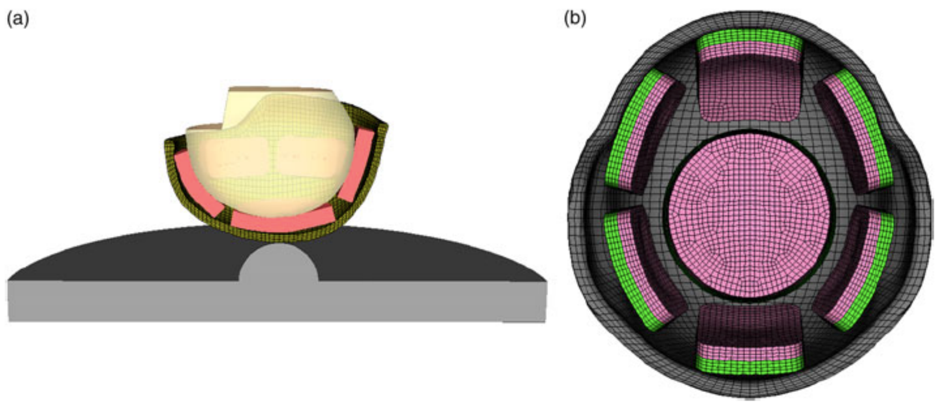
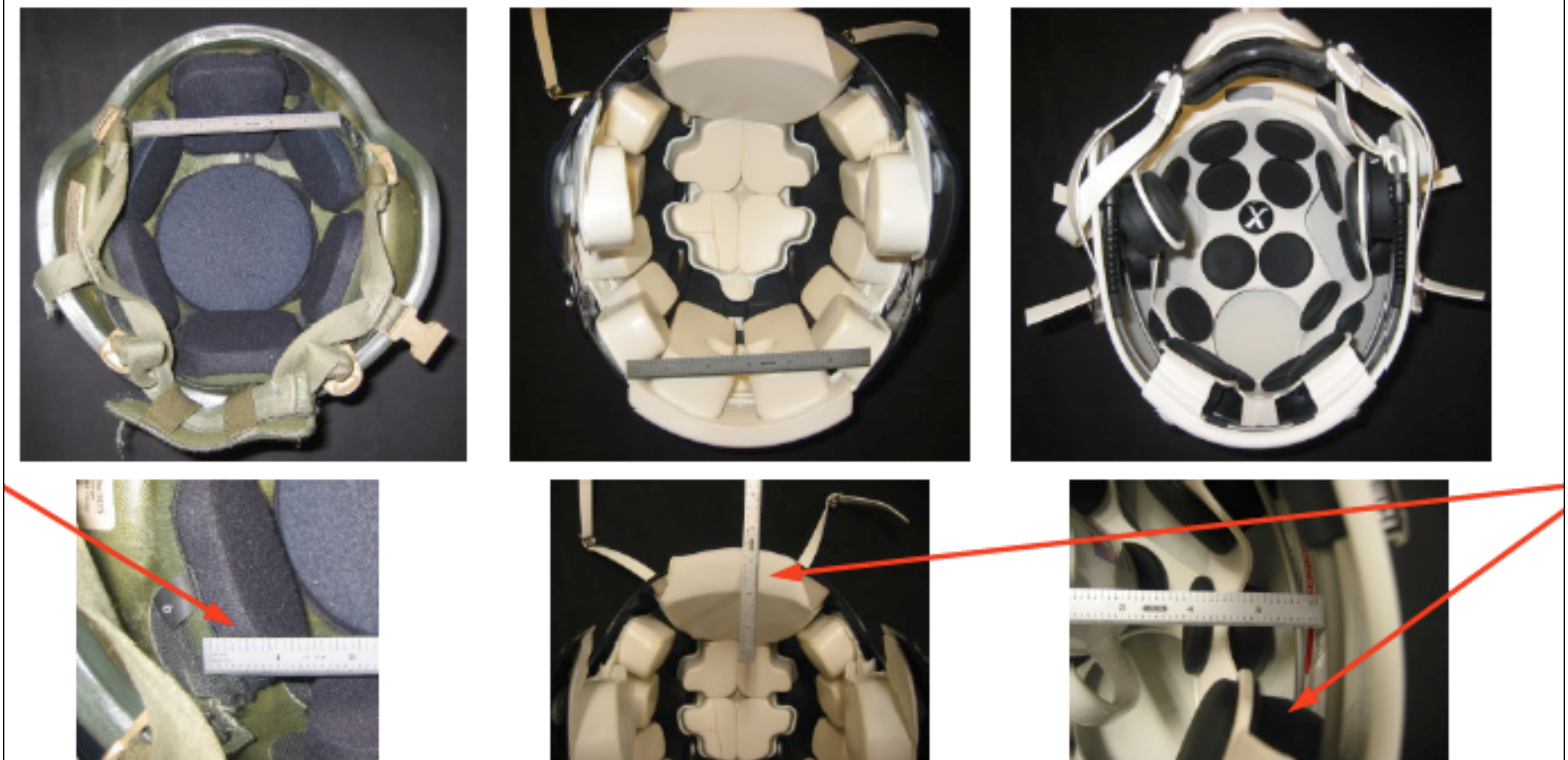
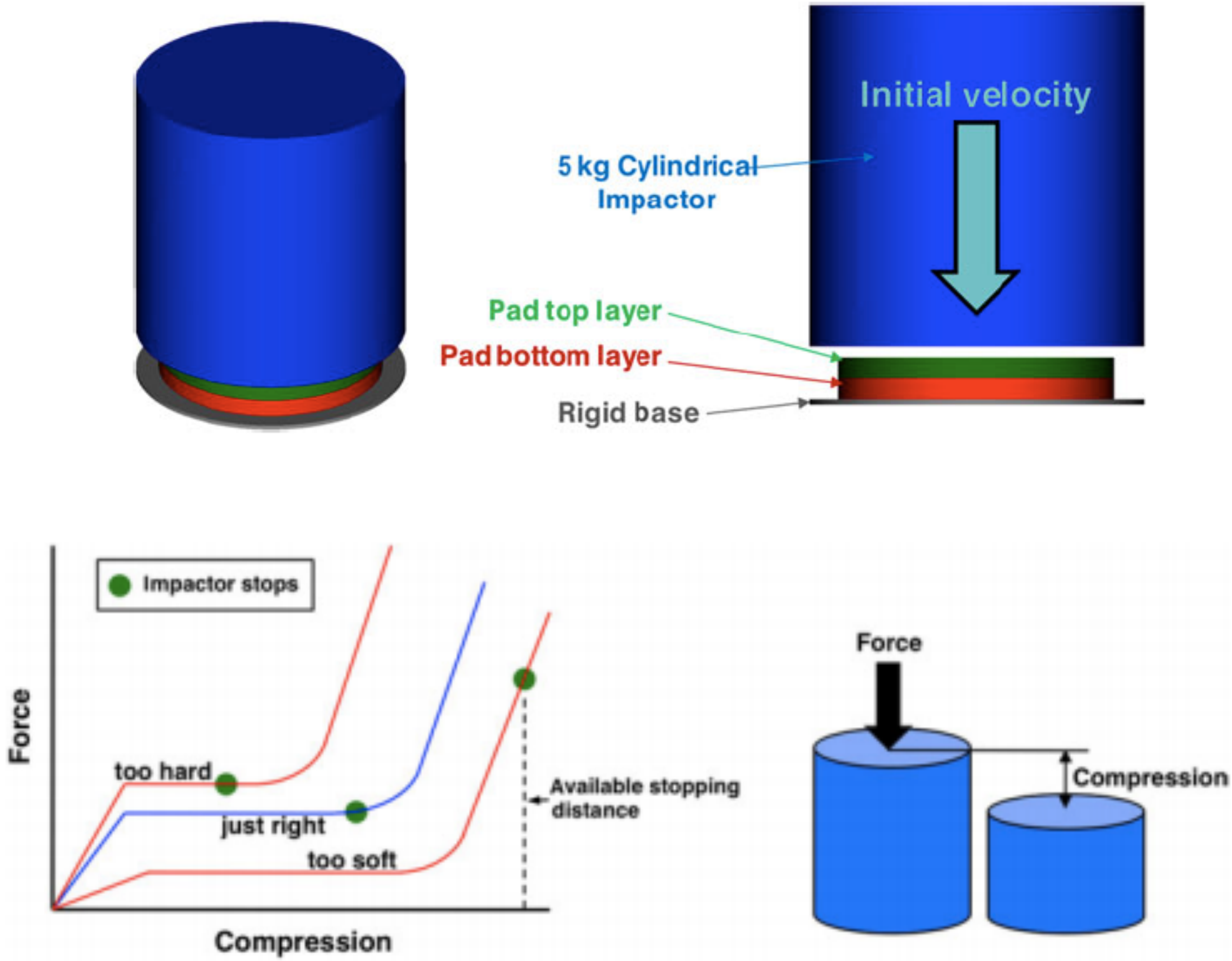


Figure 3. Geometry used to simulate USAARL experiment shown in Figure 2. (a) The half-model that was used for the simulations consists of 60,000 elements. Zoning in the helmet is shown. The headform is transparent in order to view the side pads. (b) Interior view of the complete helmet and pads.

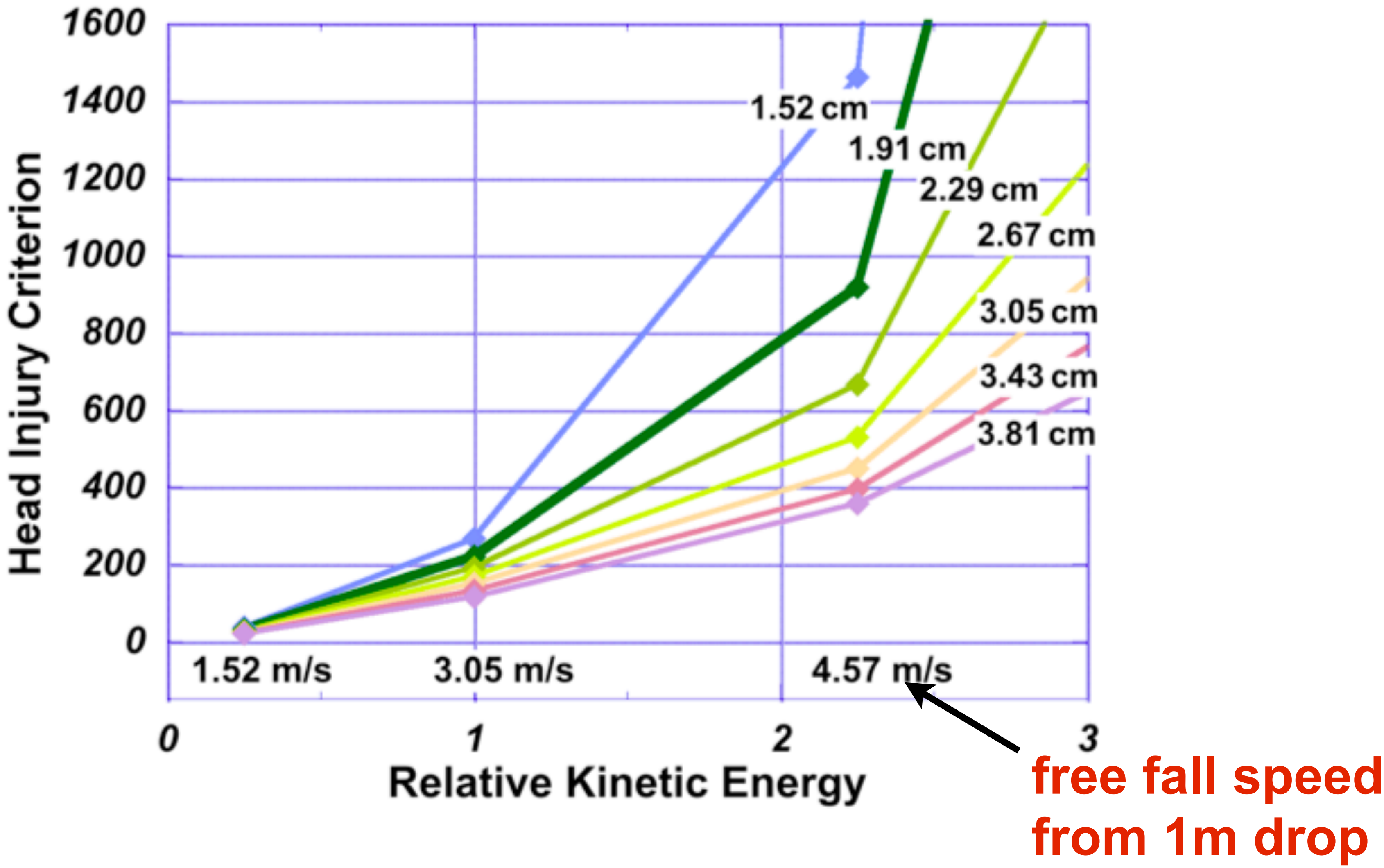
- First: 3-D Modeling of helmet drop experiments for validaton of PARADYN simulations;
- lab experiments to determine foam properties and inform viscoelastic model
- x-ray tomography to determine geometry
- meshed with help of VISIT
- then isolate pads and simulate drop of 5kg weight on isolated pads of different types



ACH **Riddell** **Xenith**



General response of foam to compression during an impact. It is assumed that the integral under each curve to the point
tor stops (green dot) is equivalent to the initial kinetic energy of the impactor.



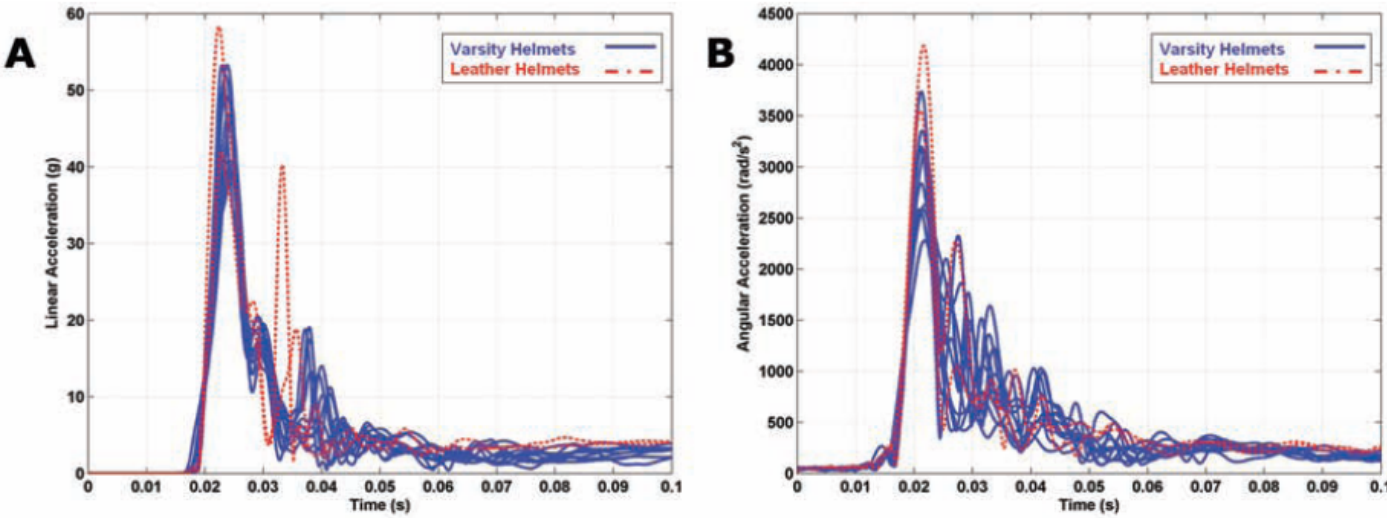
USE 1 SIZE LARGE HELMET AND FILL EXTRA SPACE WITH CONVENTIONAL PADS, INJURY REDUCTION FROM SEVERE IMPACTS IS LIKELY SUBSTANTIAL: 1/17 the risk of AIS4 on Prasad & Mertz Plot

Lessons from Pad study

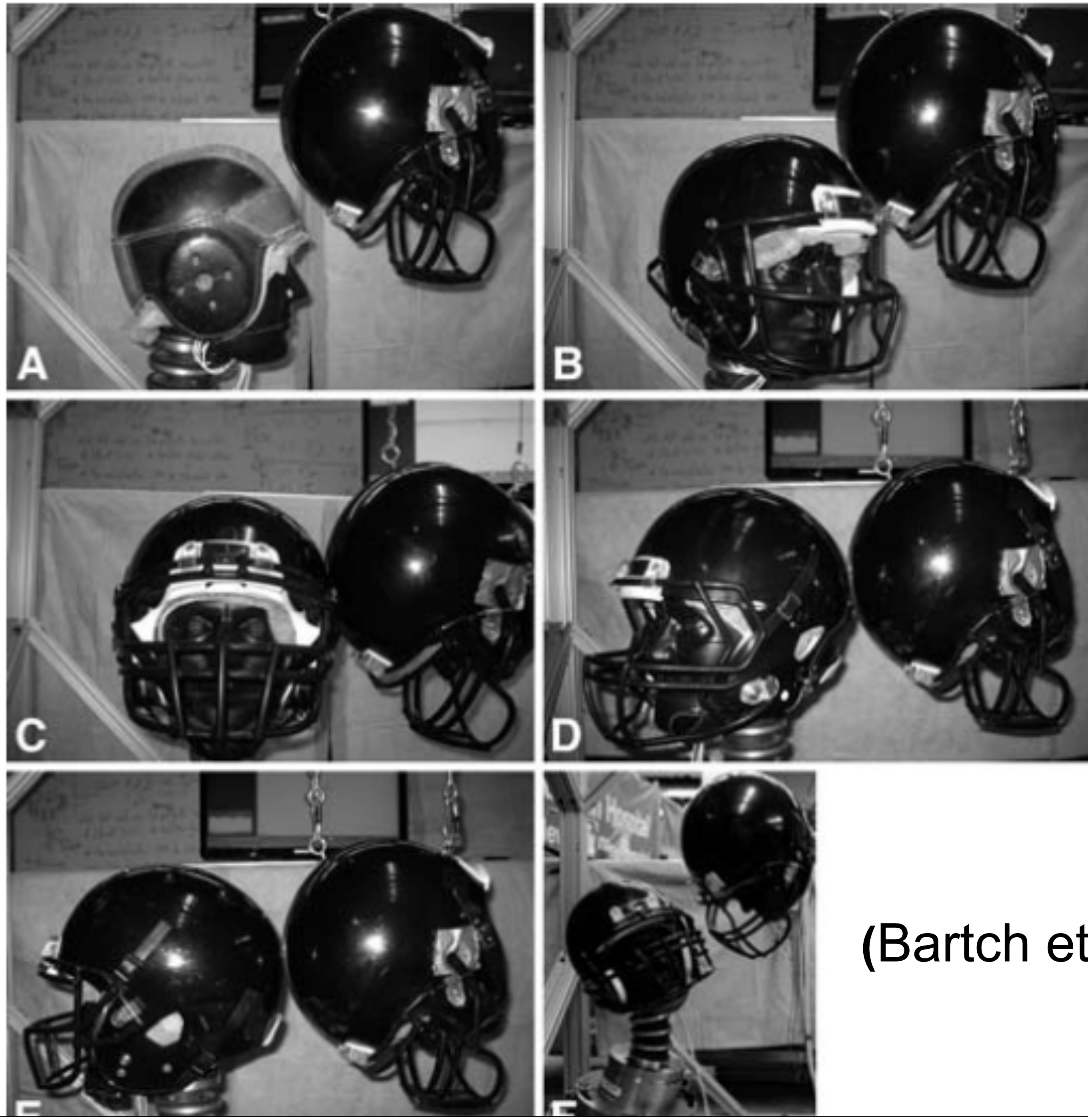
(Moss et al. 2012)

- Harder pads perform better at higher impact velocities
- Softer pads perform better at lower impact velocities
- Reducing pad area is equivalent to making pads softer
- **BEST IMMEDIATE STRATEGY: USE LARGER SIZE HELMET AND DOUBLE PADDING THICKESS**

Modern Helmets do not outperform Leather Helmets from from 1930s for subconcussive impacts (Bartch et al. 2011)



- Did not test concussive blows, but CTE is from subconcussive blows

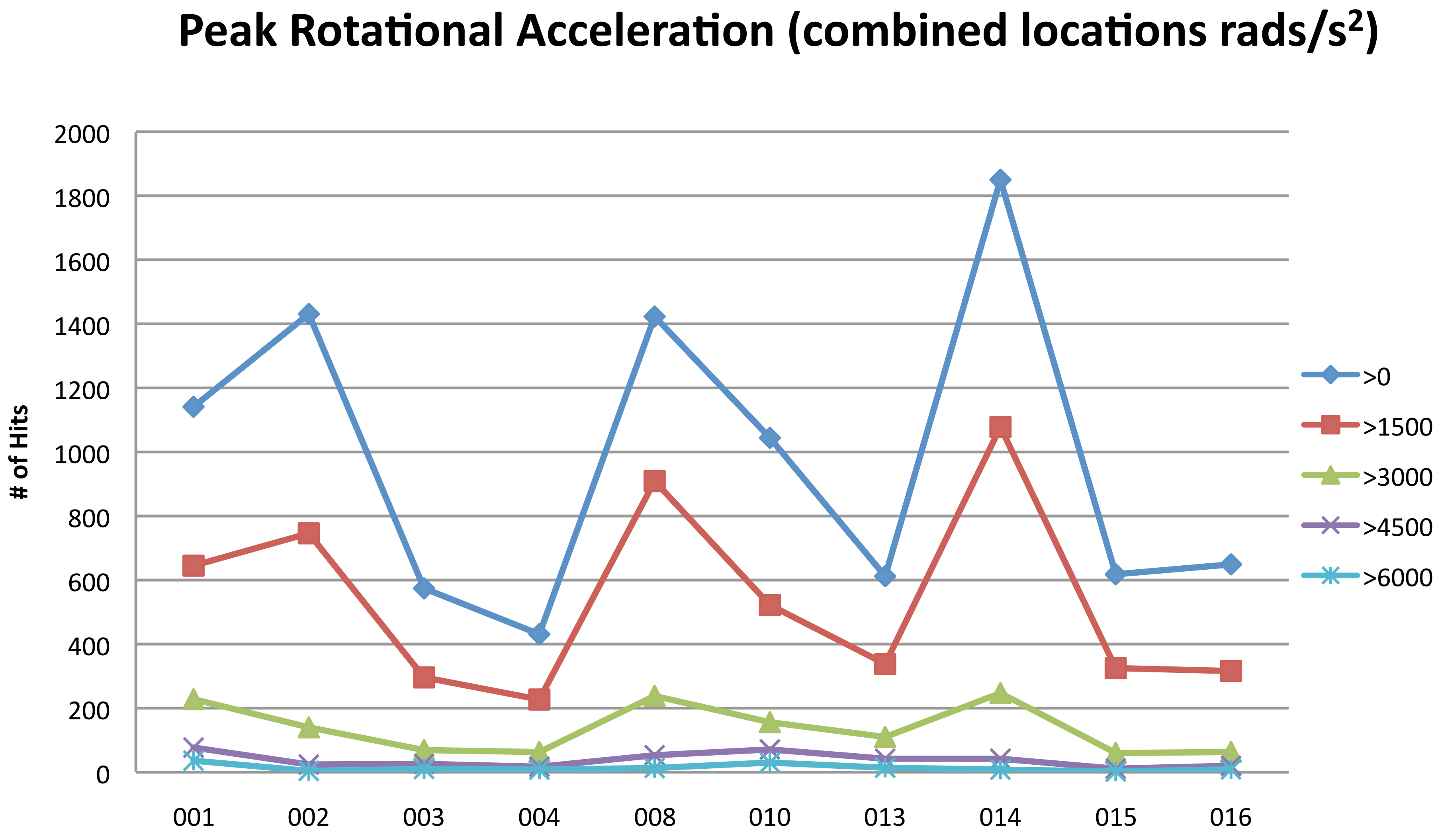


(Bartch et al. 2011)

Correlating Impact Metrics with Brain Image Changes for Sub Concussive Hits (our UR study; Bazarian et al. 2014)

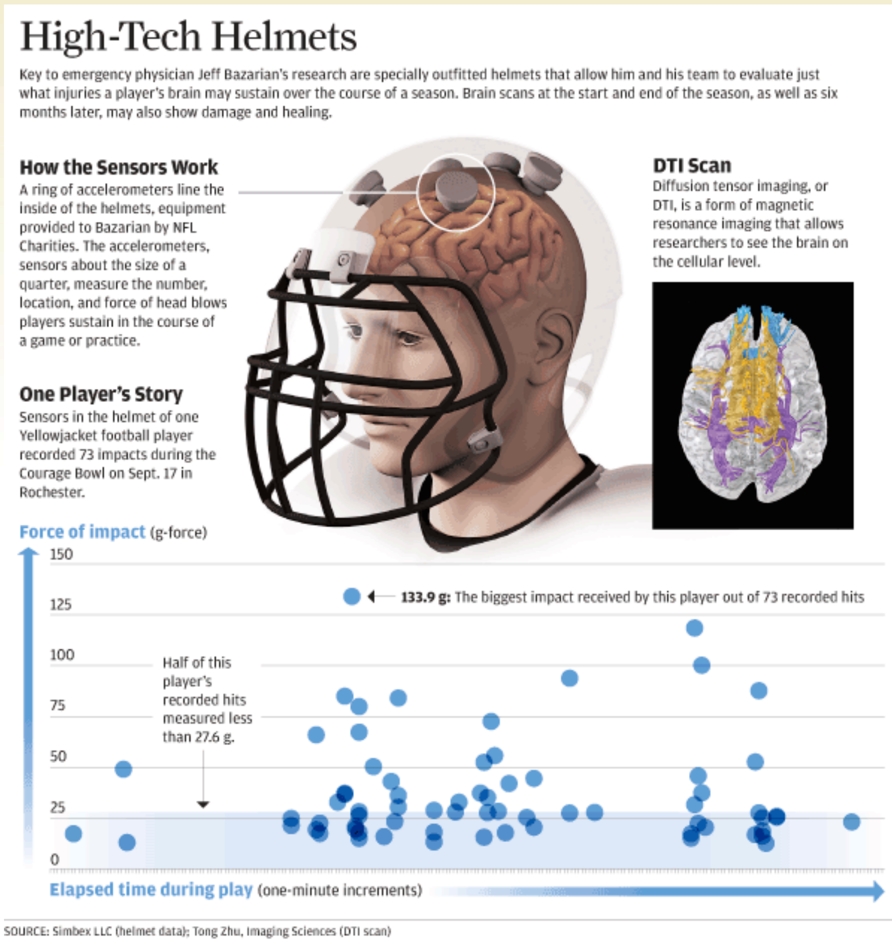
- For 10 players, accelerometers recorded mechanical data from which mechanical quantities (HIC, SI, peak g, peak rotational acceleration) were computed for each impact over the season (~ 1000 impacts per player)
- For each player, the data recorded for each quantity was binned into strength categories which became separate metrics. (e.g. the number of impacts recorded above 4500 rad/sec^2)
- Correlations between these metrics and brain image (DTI) changes in players' brain scans before and after the season were sought
- Best correlations were found with number of hits above rotational acceleraton of 4500 rad^2/sec

Example (peak rot accel.)



DTI after sub-concussive head blows

Bazarian et al. 2014



High-Tech Helmets

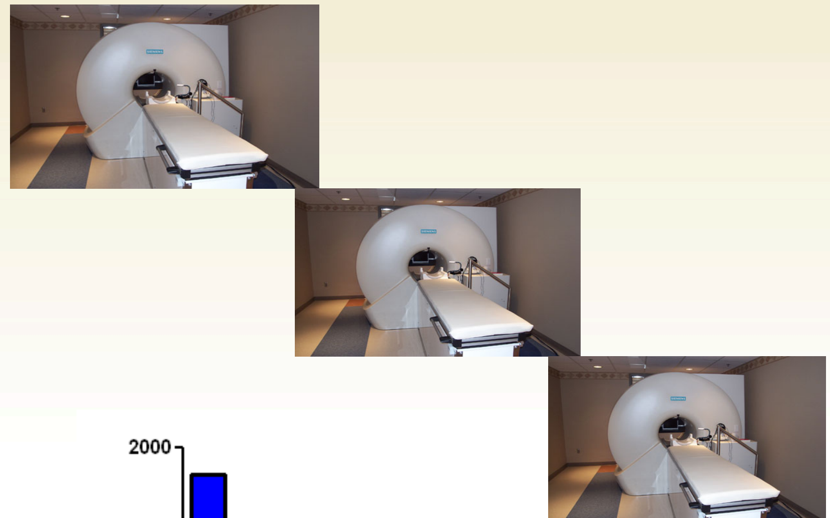
Key to emergency physician Jeff Bazarian's research are specially outfitted helmets that allow him and his team to evaluate just what injuries a player's brain may sustain over the course of a season. Brain scans at the start and end of the season, as well as six months later, may also show damage and healing.

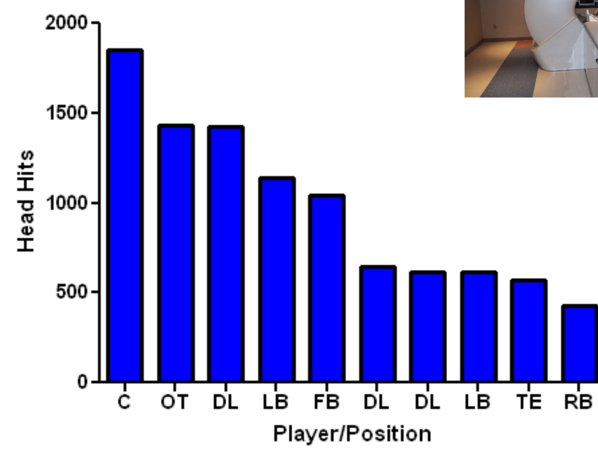
How the Sensors Work
A ring of accelerometers line the inside of the helmets, equipment provided to Bazarian by NFL Charities. The accelerometers, sensors about the size of a quarter, measure the number, location, and force of head blows players sustain in the course of a game or practice.

One Player's Story
Sensors in the helmet of one Yellowjacket football player recorded 73 impacts during the Courage Bowl on Sept. 17 in Rochester.

DTI Scan
Diffusion tensor imaging, or DTI, is a form of magnetic resonance imaging that allows researchers to see the brain on the cellular level.

SOURCE: Simbex LLC (helmet data); Tong Zhu, Imaging Sciences (DTI scan)





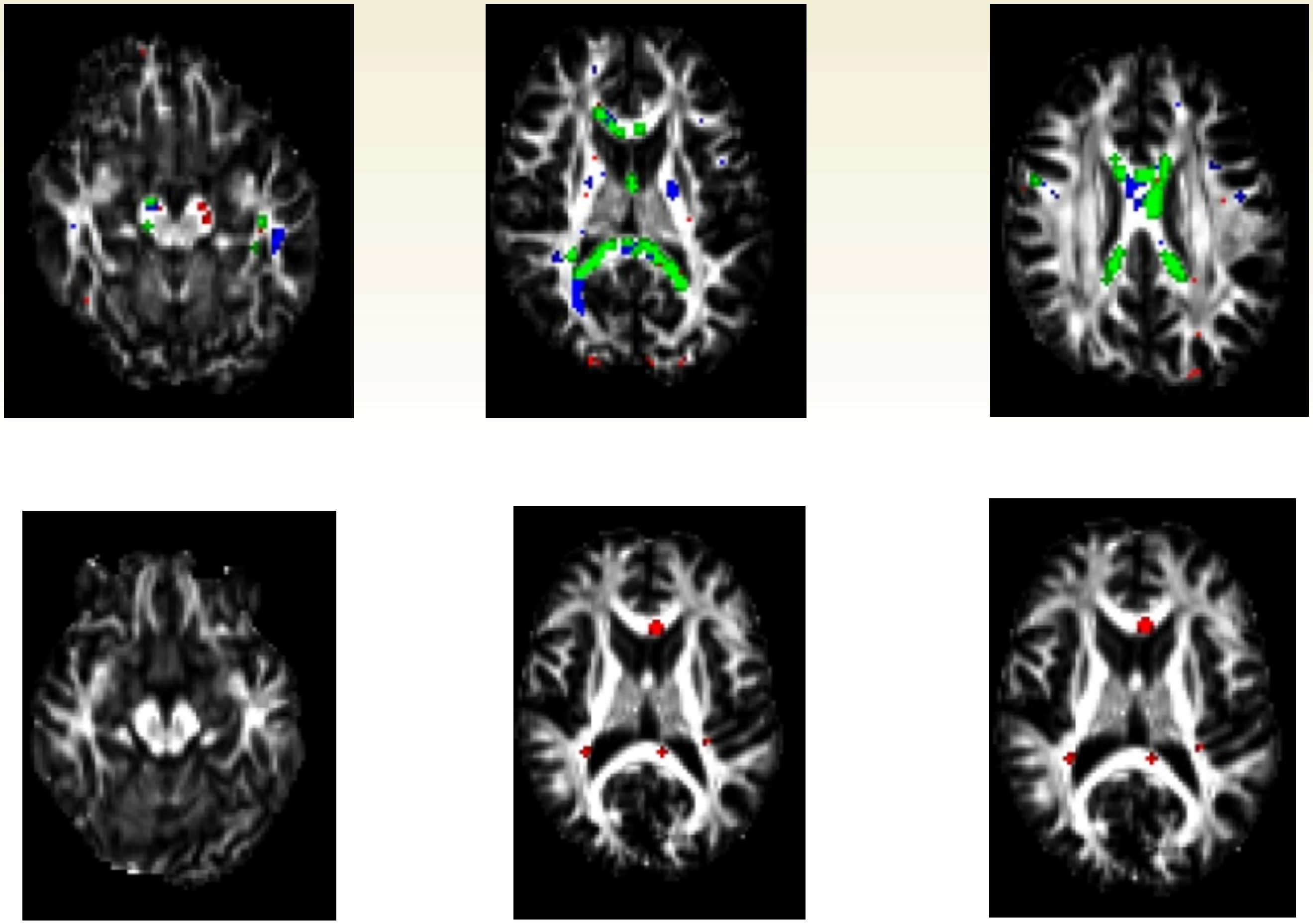
Unpublished

Illustration: Steve Boerner for Rochester Review

Co-localization of FA and MD Changes

From Bazarian et al. 2014

(T. Zhu, and J. Zhong led the DTI component of the study)



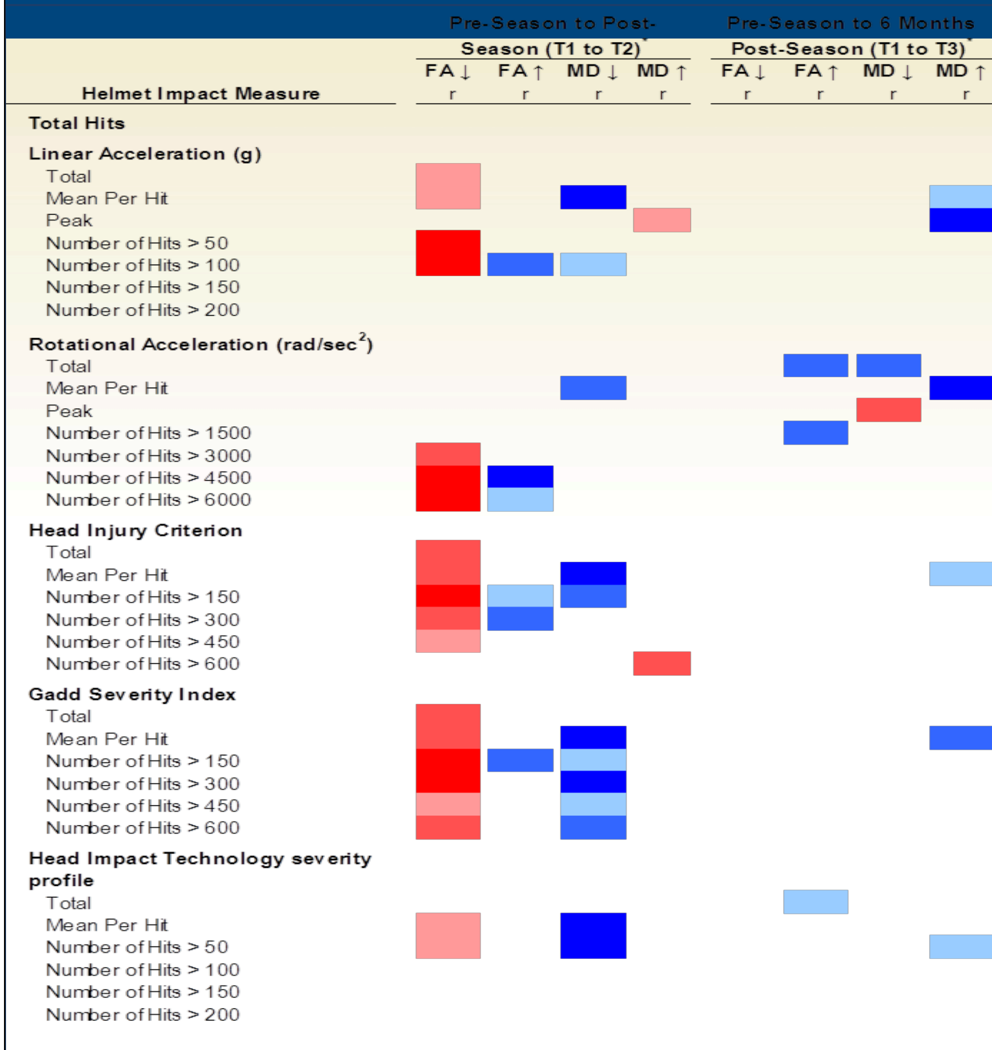
Athlete

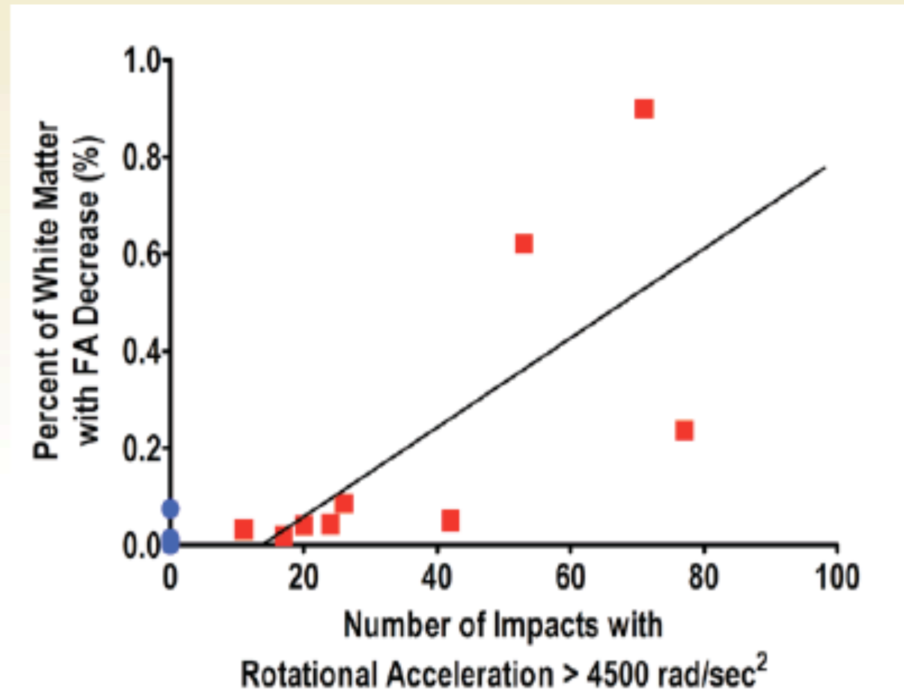
Control

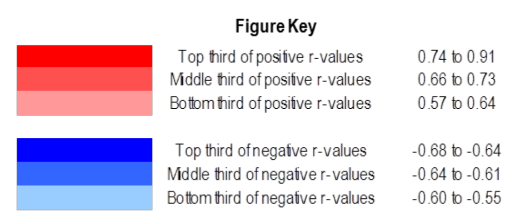
FA Change only=blue
MD Change only=red
Both FA and MD change=green

Relationship of DTI Changes to Helmet Impacts

From Bazarian et al. 2014







Importance of Mitigating Rotational Acceleration

- UR SCH study (correlation 4500 rad/s^2 with DTI changes)
- Shear moduli 50kPa (0.5 bar) << Bulk moduli 1 Gpa (10^4 bar) in brain (Horgan 2005)
- rotational acceleration Q in rad/s^2 corresponds to tangential linear acceleration Q/100 in g: *e.g*. 1000 rad/s^2 ~ 10 g
- empirical threshold accelerations for concussions are lower using the above conversion
- current helmets do little to mitigate rotational acceleration; need decoupled shell
- mice subjected to blast wave; 10^6 rad/s^2; showed CTE, 2 weeks after blast but not when head was fixed (Goldstein et al. 2012)
- might blast injury also be related to rotational accel?

NFL MTBI measures (King 03; Zhang et al. 04; Newman 05)

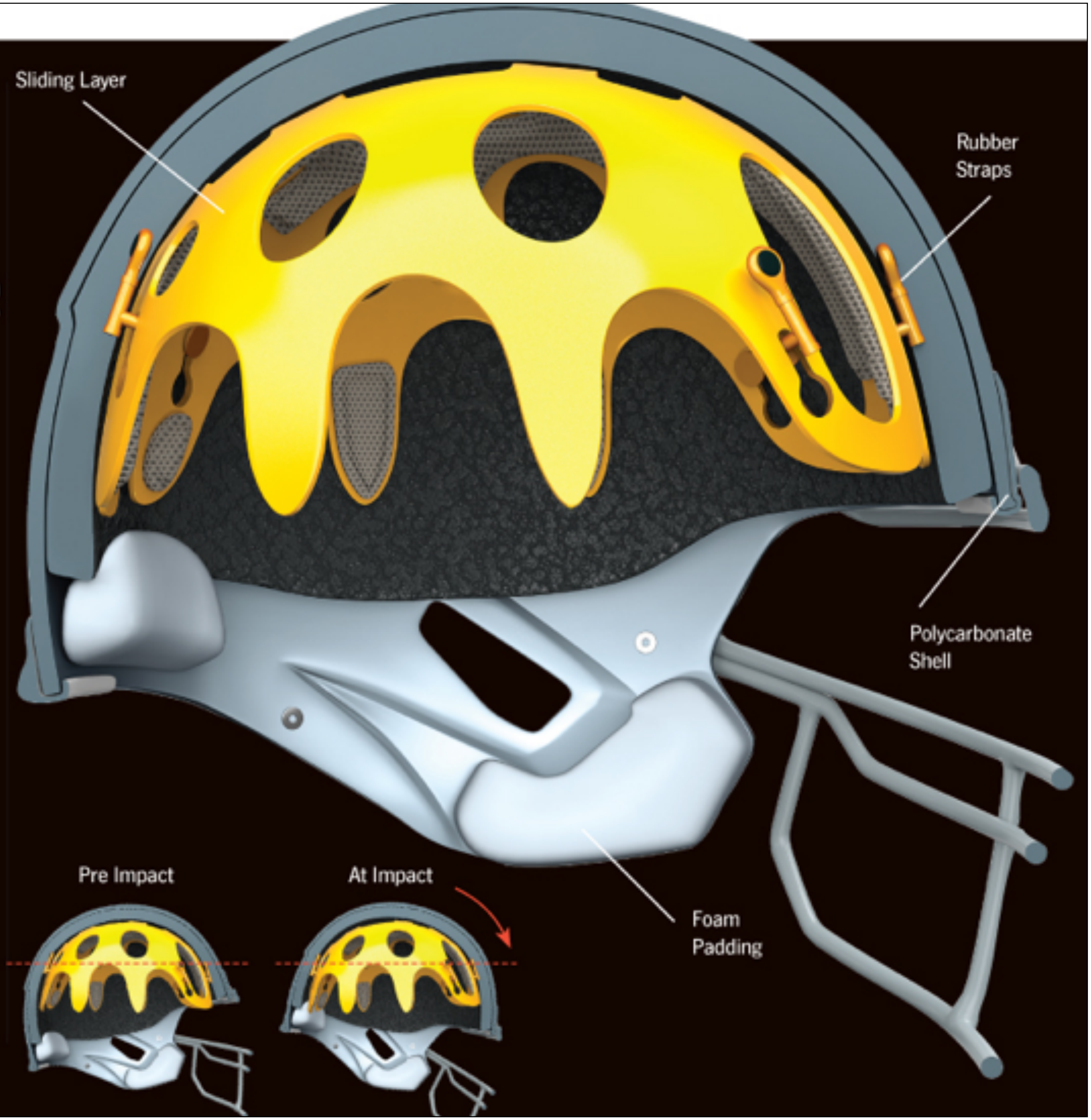
Predictor Variable	Threshold Values for Likelihood of MTBI		
	25%	50%	75%
$A_{r\ max}$ (m/s^2)	559	778	965
$R_{r\ max}$ (rad/s^2)	4384	5757	7130
HIC_{15}	136	235	333
ε_{max}	0.25	0.37	0.49
$d\varepsilon/dt_{max}$ (s^{-1})	46	60	79
$\varepsilon \bullet d\varepsilon/dt_{max}$ (s^{-1})	14	20	25
Stress at midbrain (brain stem)	6 kPa 0.06 atm	8 kPa 0.08 atm	10 kPa 0.1 atm

The Helmet that Might Save Football

THE SYSTEM The Multidirectional Impact Protection System (MIPS) reduces the rotational forces that cause concussions. In a MIPS-equipped helmet, a thin layer of molded plastic fits atop a player's head, beneath the padding and hard polycarbonate shell. Rubber straps affix the MIPS layer to the helmet.

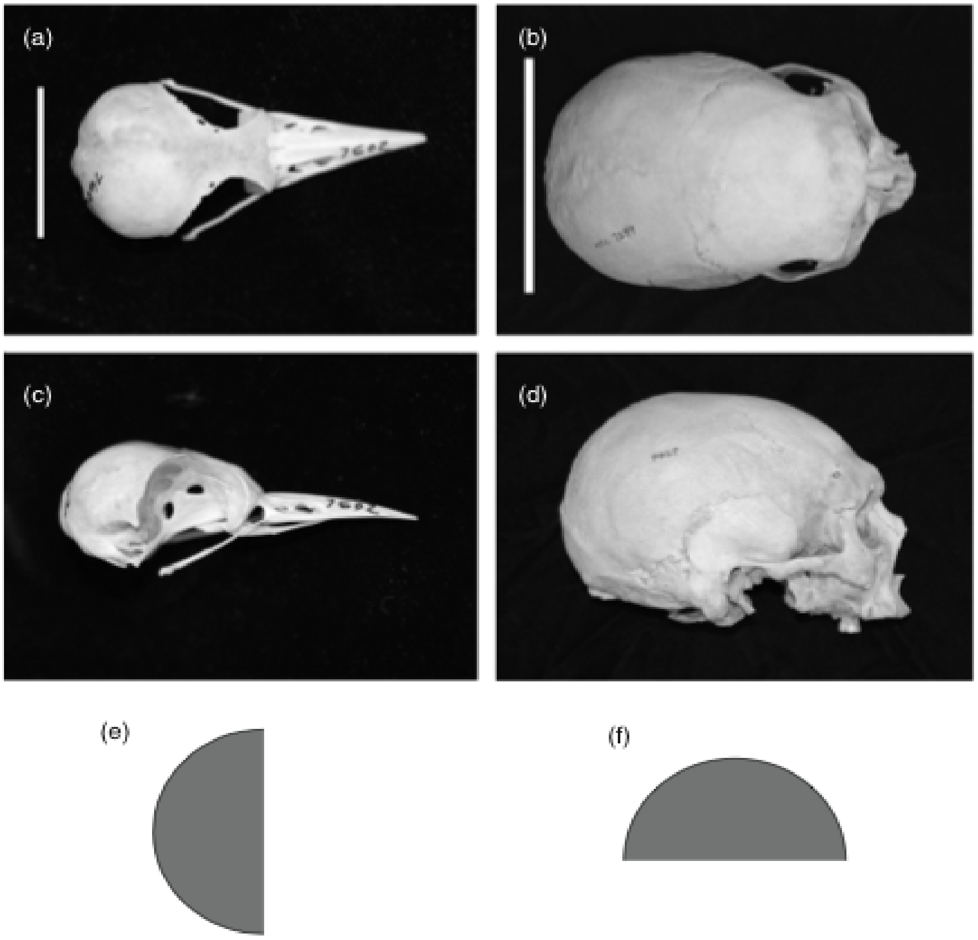
HOW IT WORKS MIPS mimics the human head's own protective system, in which a layer of slippery cerebrospinal fluid sits between the brain and the skull. When an impact occurs, the skull can rotate just a bit relative to the brain. With MIPS, the rubber straps allow the helmet to move just a bit relative to the sliding, low-friction head cap, thereby eliminating much of the twisting motion before it reaches the brain.

THE RESULTS In lab tests, MIPS reduces brain rotation by as much as



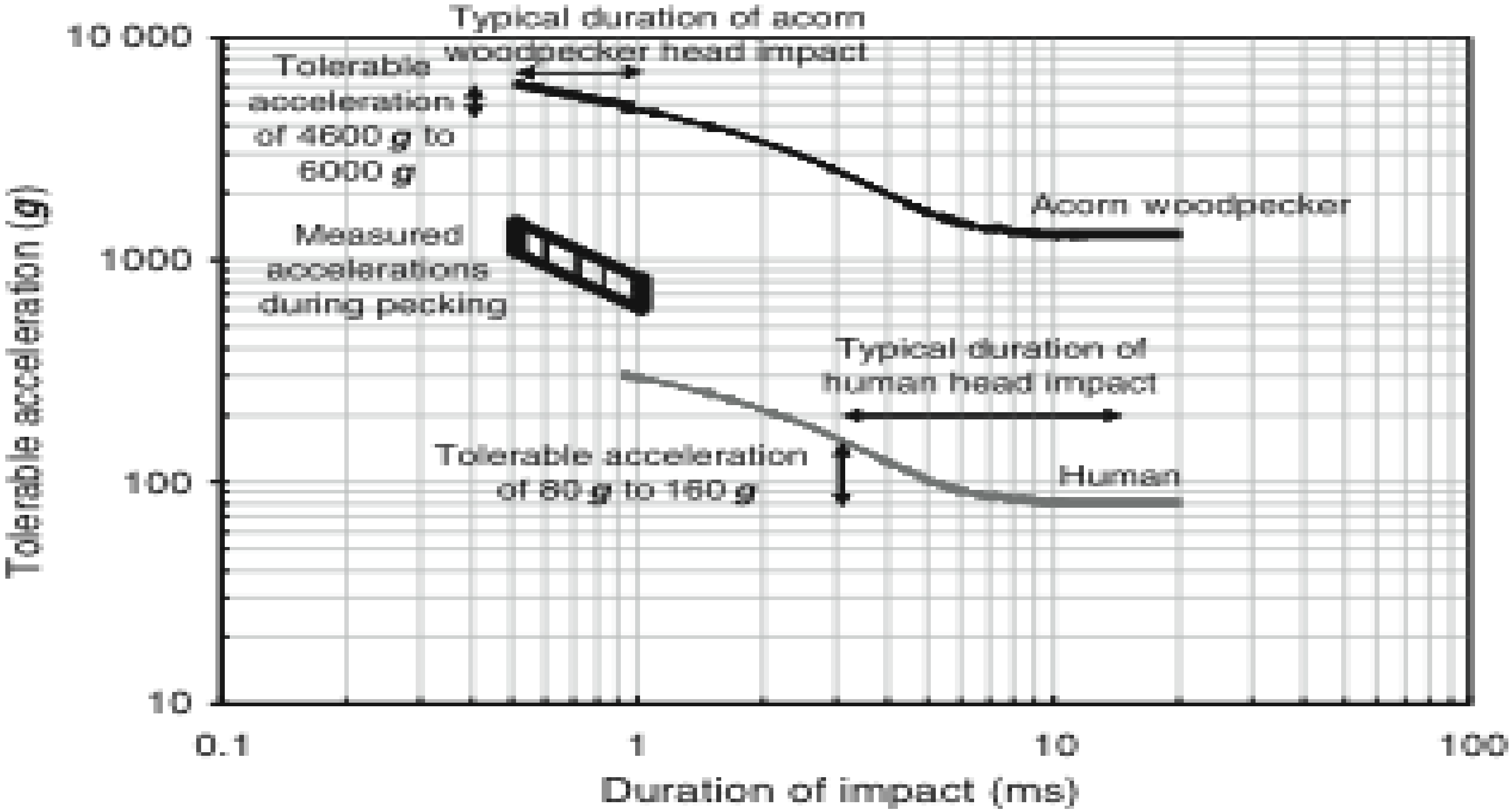
Example of the utility of "Physics" thinking:

- Woodpecker head accelerations are ~1000g, with HIC > 300,000 sec; humans get severe TBI at HIC >1000
- **Why don't woodpeckers get TBI? :**



Gibson (2006)

- Standard TBI metrics rely on fixed brain mass and surface area
- for similar brain tissue, woodpecker impacts, when corrected for brain size and orientation are below all reasonable injury thresholds for concussions



More on Woodpeckers....

- Woodpeckers also have tighter fit between brain and skull--less sloshing OK because of lighter brain!!
- spongy skull (Wang et al 2011)
- very little rotational acceleration (Wang et al 2011)
- **Not just about concussions but also protection against subconcussive impacts/CTE: woodpeckers likely have features that prevent BOTH concussions and CTE damage**

Numerically Simulating Impact + Head Models: TBI Thresholds based on Internal Forces

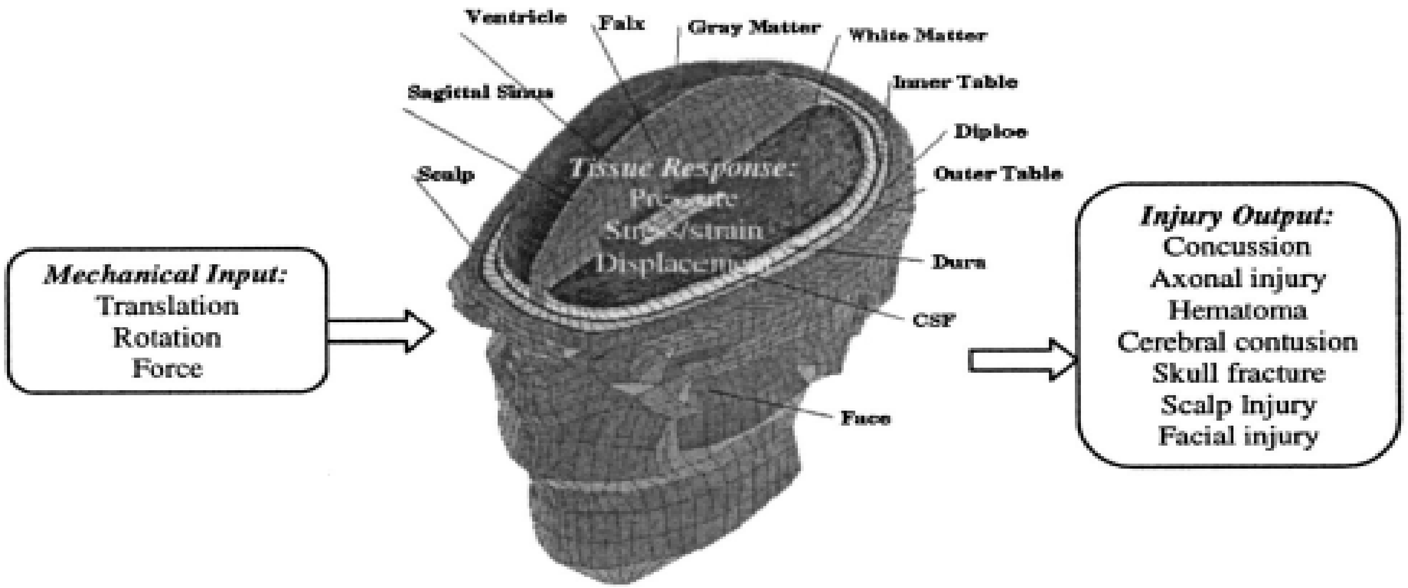
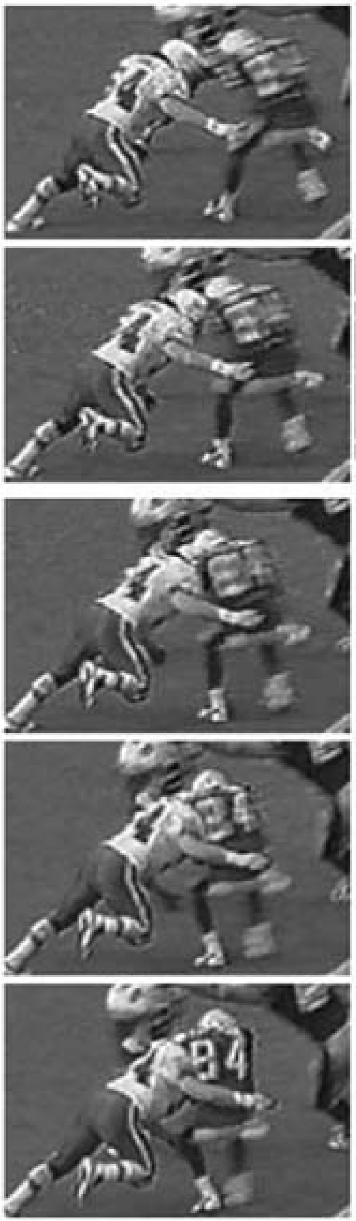


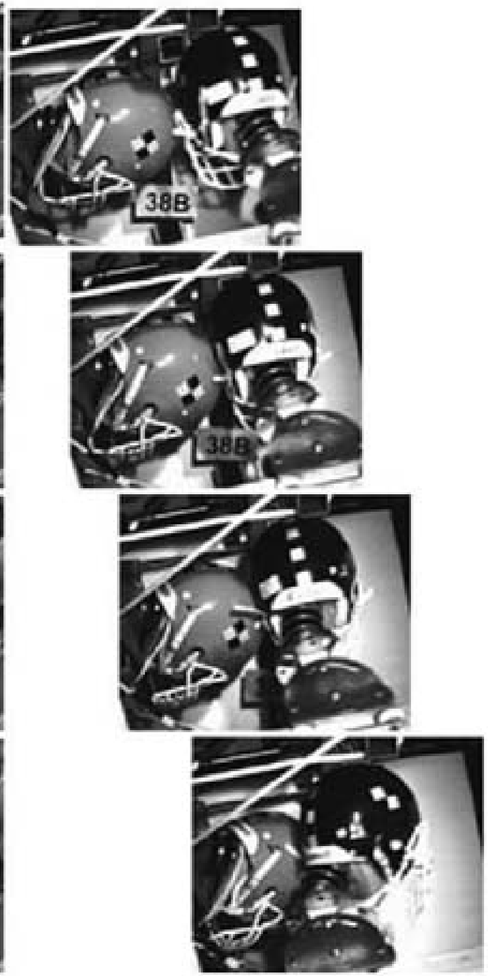
Fig. 2 Block diagram illustrating the tissue-injury response to traumatic input loading as a biomechanical analysis process using a computer surrogate

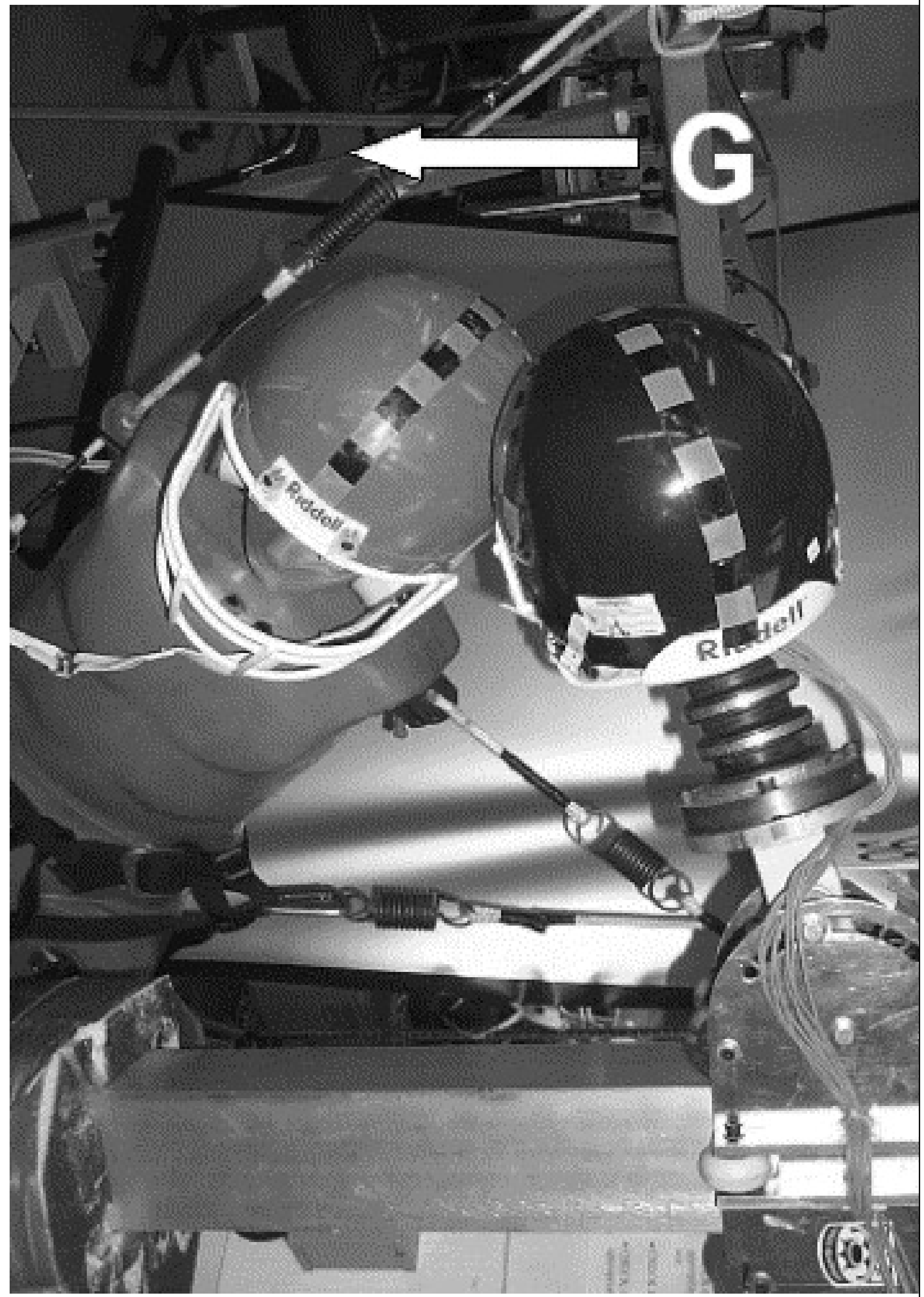
- Zhang et al. 04; reproduced NFL collisions with Wayne State Head Model
- WSHM: gray matter (cell) white (fibrous): shear moduli 20% larger for white; white is 2-D isotropic, grey is 3-D isotropic; brain stem shear mod 40% higher than cerebrum etc..
- Data on these properties differ, but code can incorporate whatever the data require

Viano et al 2007

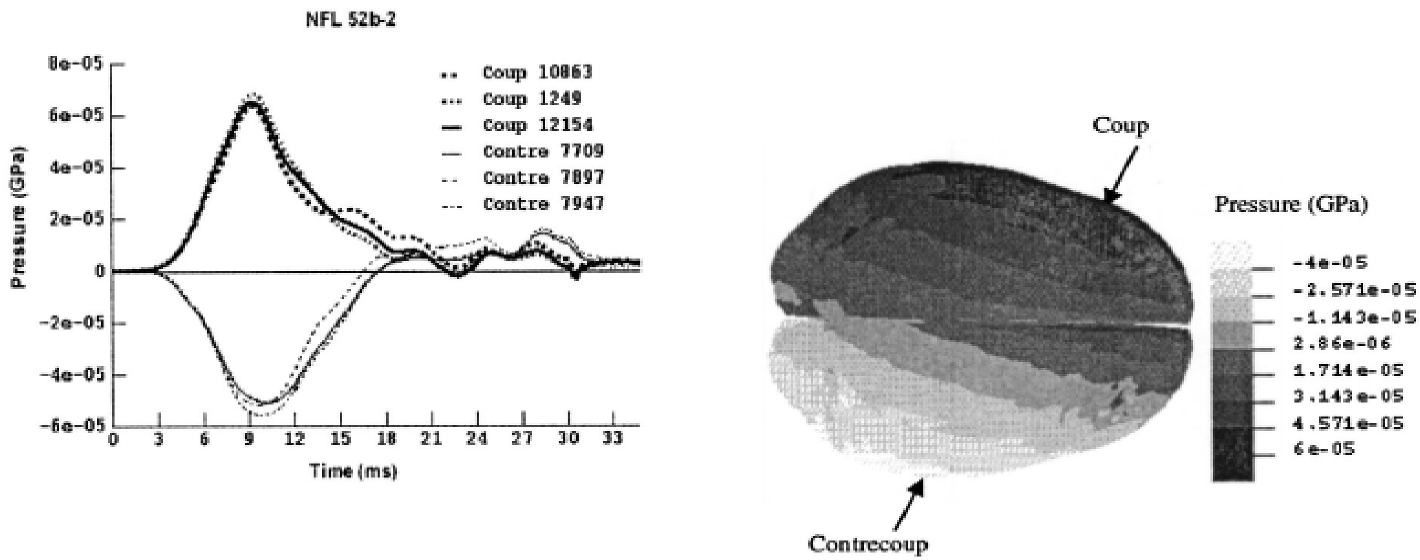


(Pellman et al. 03)





Coup + Contrecoup pressures



(A) (B)

Fig. 4 (*A*) Predicted peak positive and peak negative intracranial pressure-time histories; (*B*) Predicted intracranial pressure distribution 9 ms after the impact

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Liyiang et al (04)

TBI “internal” measures from simulations

- Zhang et al 04: reproduced game video impacts with head forms in lab, then use lab data as input for numerical simulations to calculate internal stresses
- Maximum stress at core (diencephalon, upper brain stem)
- rate of maximum strain (= rate of elastic energy change) and peak stress are best correlators with concussive injury
-

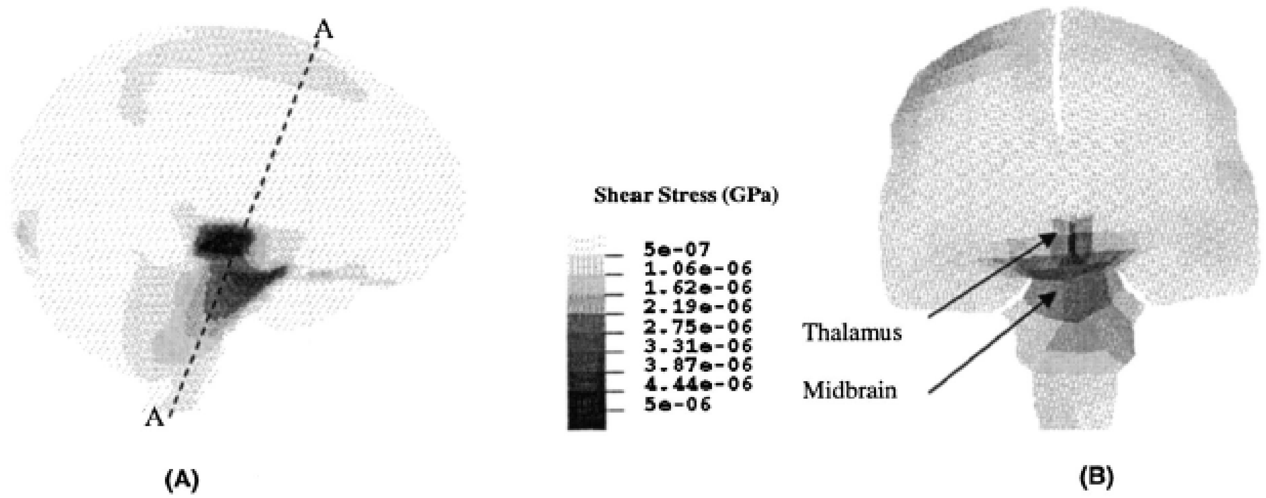
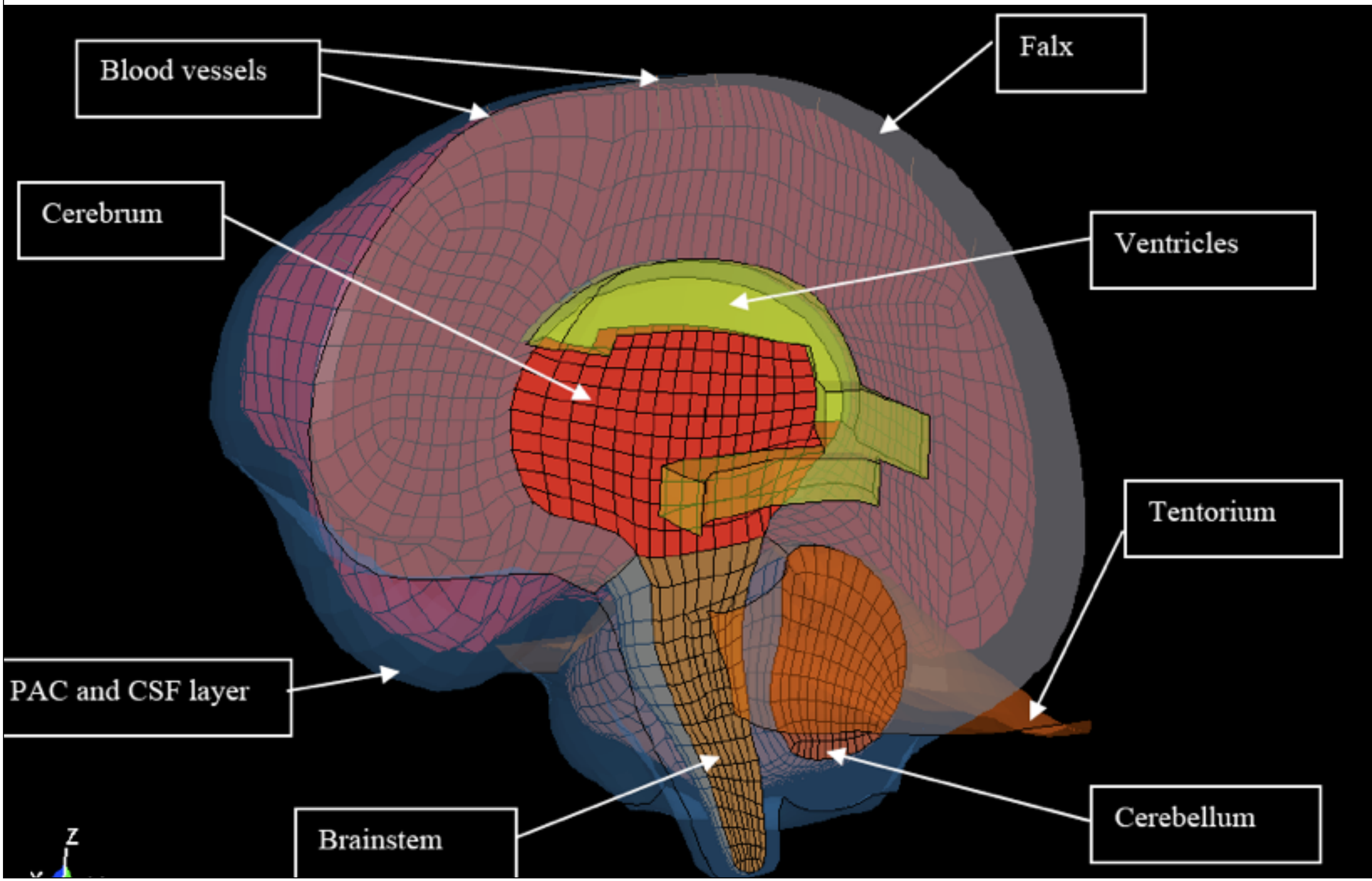


Fig. 6 Shear stress contours predicted by the model at 16 ms for: (*A*) a parasagittal section; (*B*) Sectional view through A-A which represents a section through the mamilothalamic tract

SIMon (Simulated Injury Montor FE model) Takhounts et al. 2008) NHTSA



Comparing Impact and Blast TBI:

Moss, King, Blackman (2009,2012)

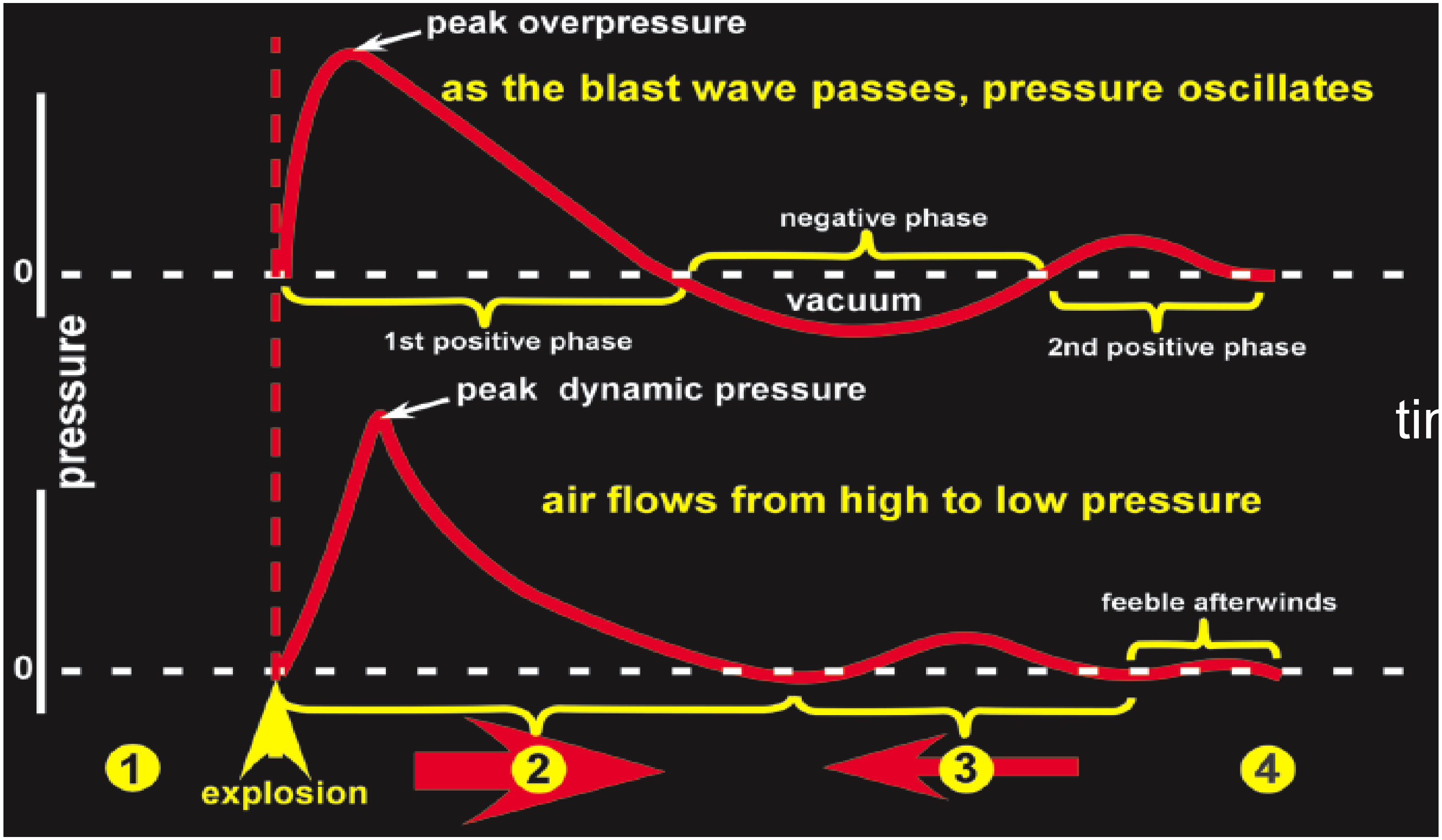
ALE3D: LLNL's blast analysis code

- Originally developed to support the nuclear weapons complex

- 3D Arbitrary Lagrangian-Eulerian Hydrocode
  - Advection capabilities
  - Built in methods for coupling fluid and structural interactions
  - Complex geometries
  - Massively parallel capabilities ← analysis with supercomputers
  - Rich material library
  - Thoroughly tested

- ALE3D is specially designed for studying the response of complex structures to blast

Basic Blast Physics



- Hopkinson's Rule:

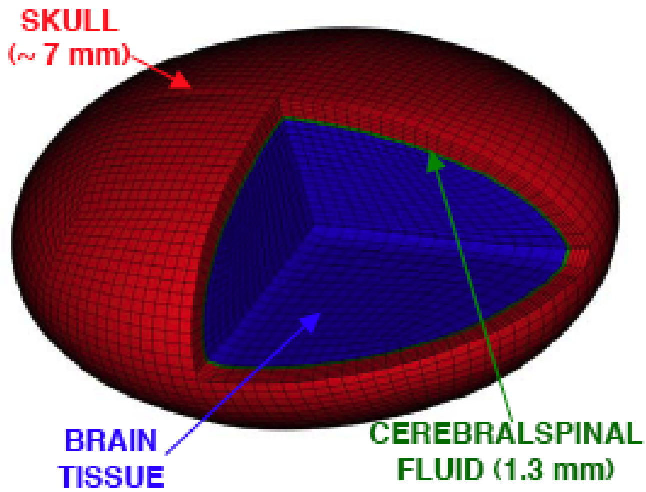
$$\text{Overpressure} \ \ P = 0.5\text{atm} \left(\left(\frac{d}{10 \text{ ft}} \right) \left(\frac{W_{TNT}}{1 \text{ lb}} \right)^{-1/3} \right)$$

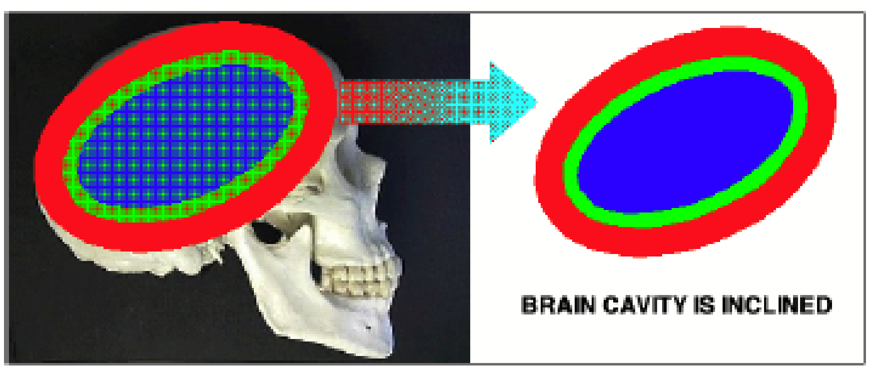
Blast Wave Injury

Blast Overpressure	Physiological Effect
0.2 psi (~0.01 atm)	Minor Ear Damage
1 psi (~0.1 atm)	Knock a Person Over
5 psi (~0.34 atm)	Eardrum Damage
15 psi (~1 atm)	Lung Damage
35 psi (~2.4 atm)	Fatalities Possible
65 psi (~4.4 atm)	Fatality Almost Certain

(Moss & King, personal comm.)

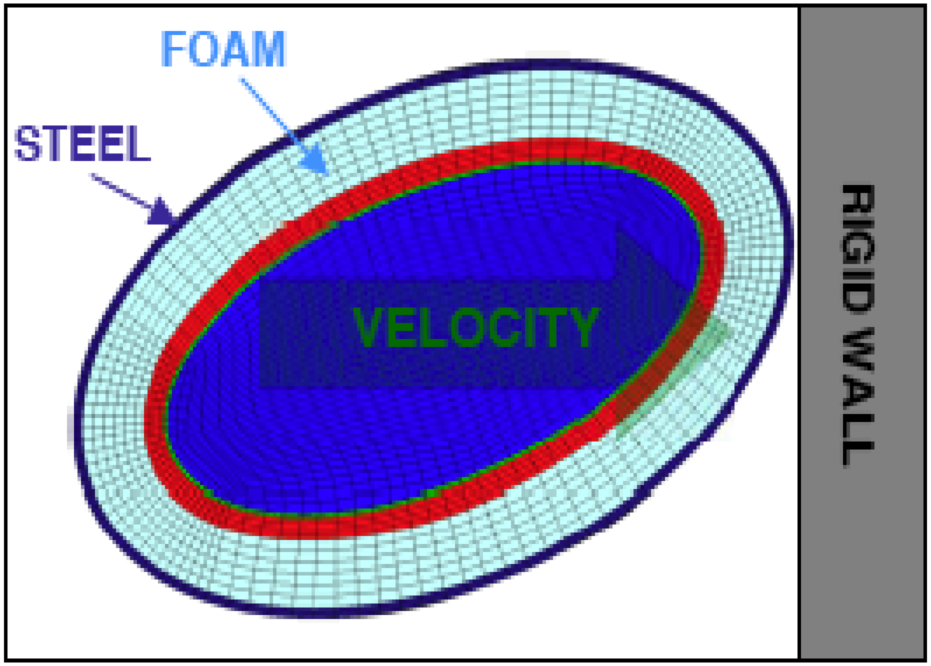
“Head” in Minimalist Simulatons





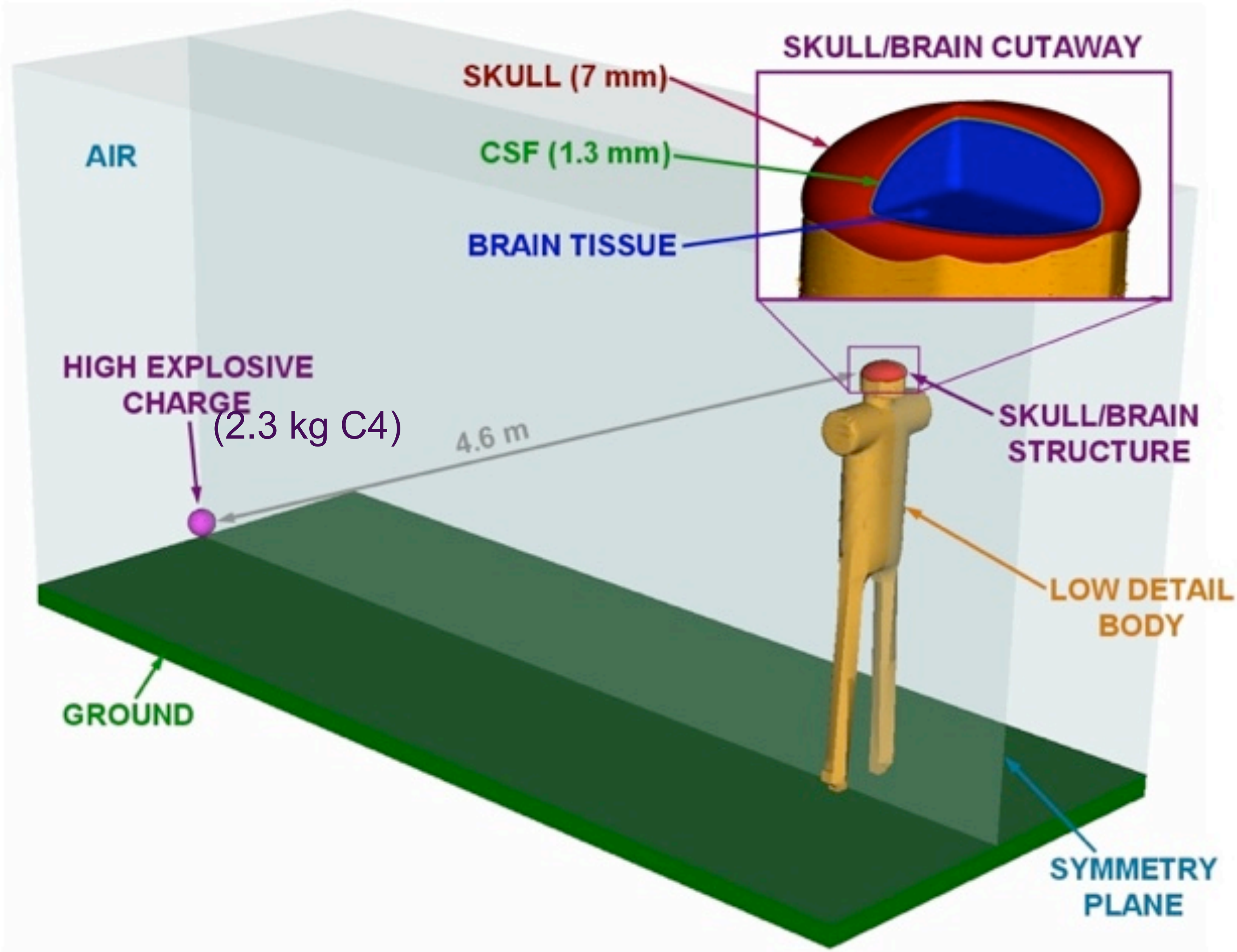
Brain cavity model

Model for Impact

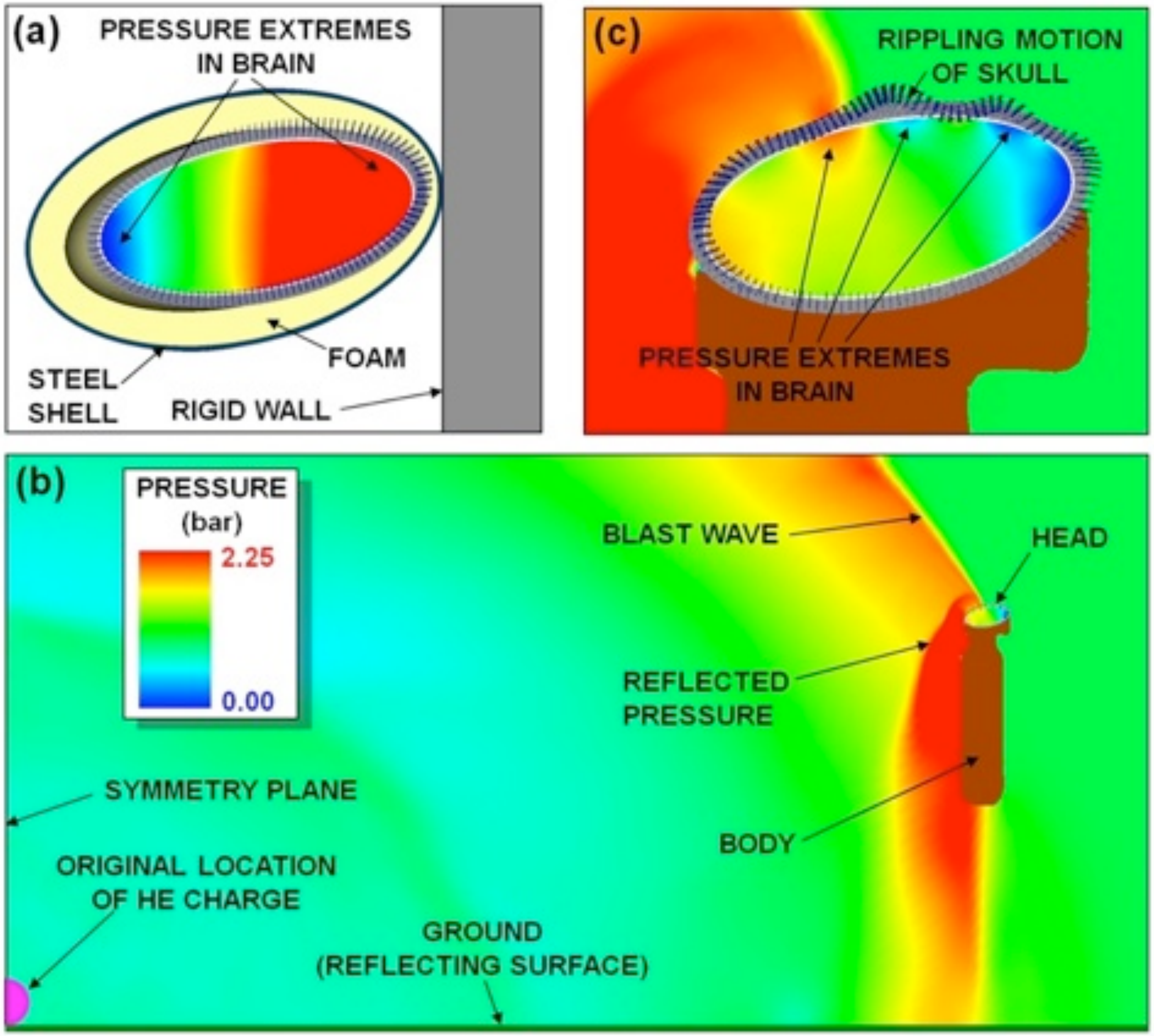


Impact model with padded helmet (angled)

- HIC = 1090
- peak g 194 g
- impact duration 2.1 ms

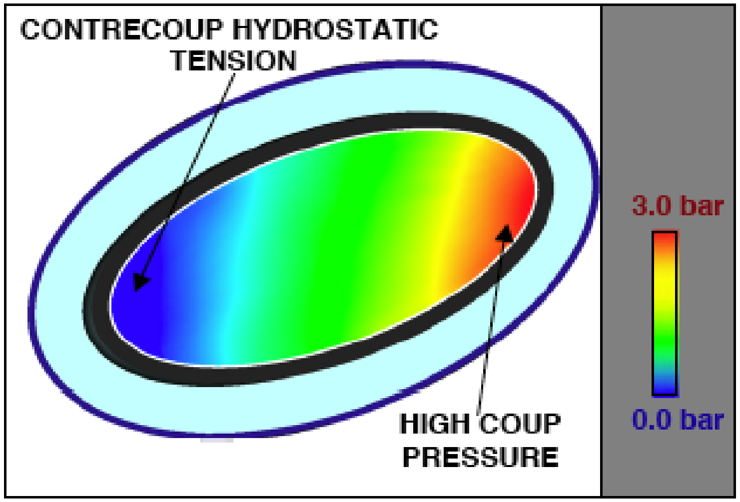

SKULL/BRAIN CUTAWAY
SKULL (7 mm)
CSF (1.3 mm)
AIR
BRAIN TISSUE
HIGH EXPLOSIVE
CHARGE
(2.3 kg C4)
4.6 m
SKULL/BRAIN
STRUCTURE
LOW DETAIL
BODY
GROUND
SYMMETRY
PLANE

Snapshot of Impact vs Blast Pressures

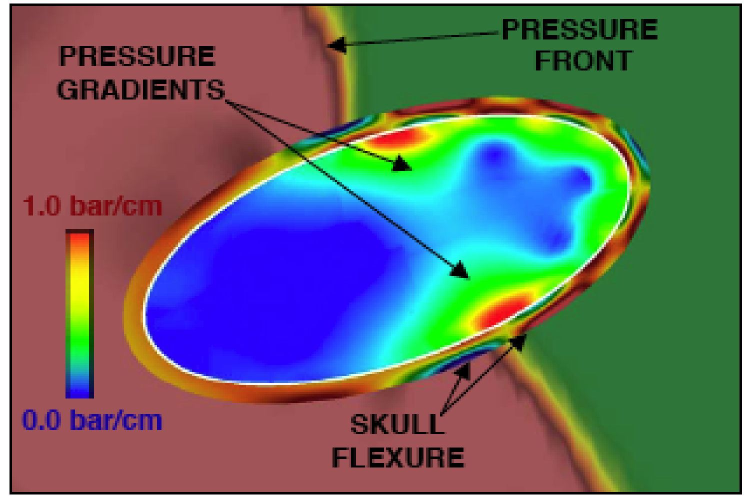


Blast wave at 5.6 ms after detonation

Impact versus blast



PRESSURES DURING IMPACT



PRESSURE GRADIENTS DURING BLAST

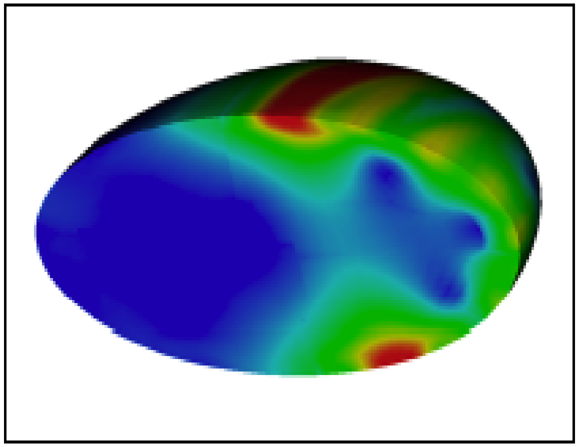
<u>Impact</u>

- Large linear accelerations
- Angled→large rotations
- Moderate skull flexure at ends
- High coup pressure
- Contrecoup tension → cavitation
- Small pressure gradients
- Rotation→large shear strains, bridging vein stretching

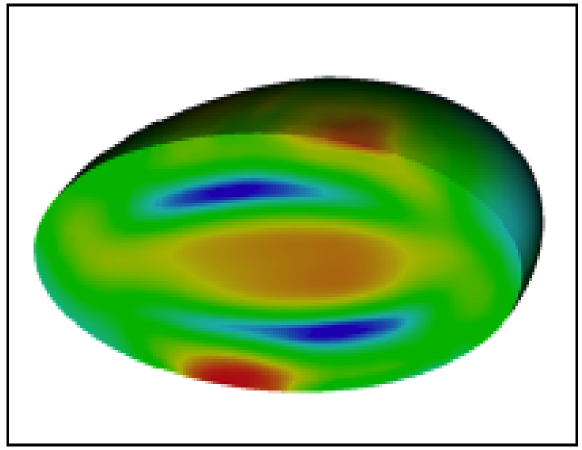
<u>Blast</u>

- Negligible linear acceleration
- Small rotations (more with whiplash?)
- Pressure wave → large lateral skull flexure
- Moderate coup/contrecoup pressure
- Hydrostatic tension → cavitation
- Skull flexure → large pressure gradients
- Rotation, pressure gradients → moderate shear strains

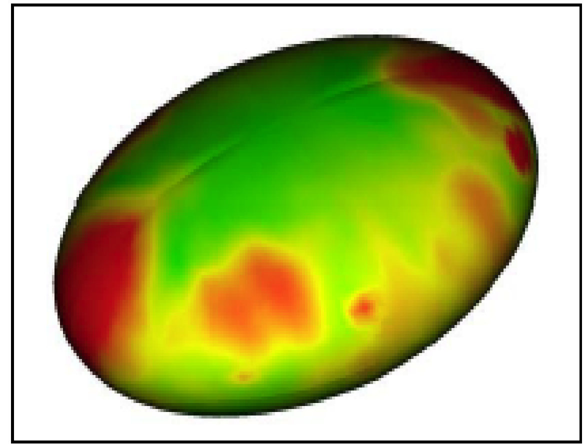
POTENTIAL DAMAGE METRICS FROM VARIOUS SIMULATIONS



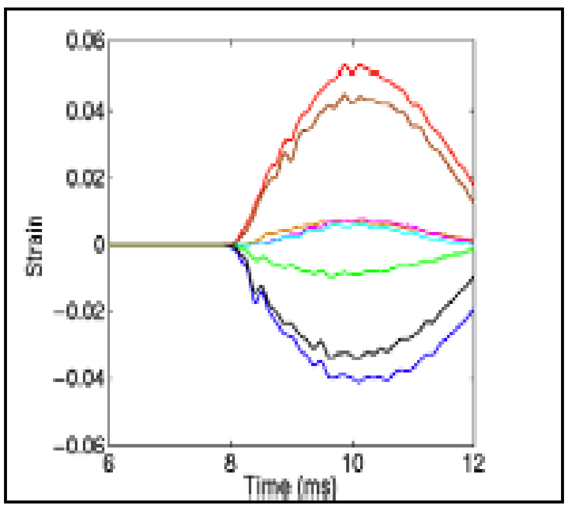
Pressure gradients in brain
HEMORRHAGING, DIFFUSE AXONAL INJURY



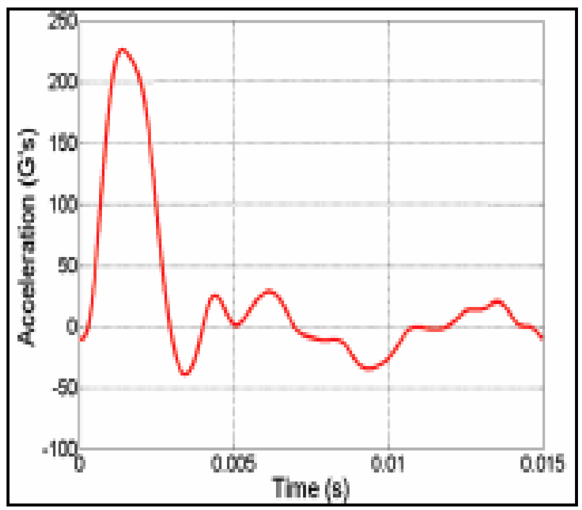
Shear strain in brain tissue
DIFFUSE AXONAL INJURY



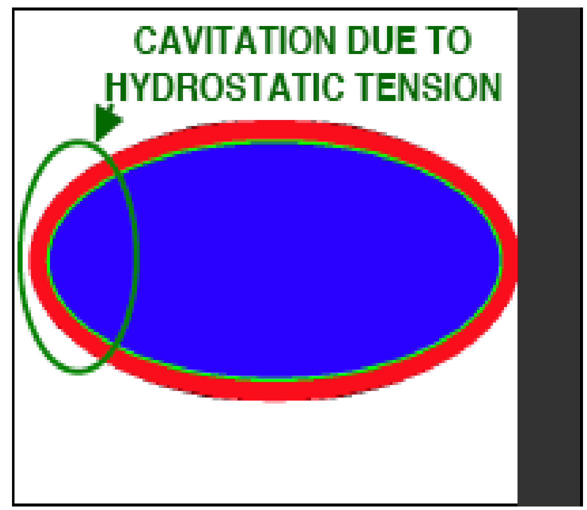
Peak pressures at brain surface
CONTUSIONS



Strain in bridging veins between brain and skull
SUBDURAL HEMATOMA



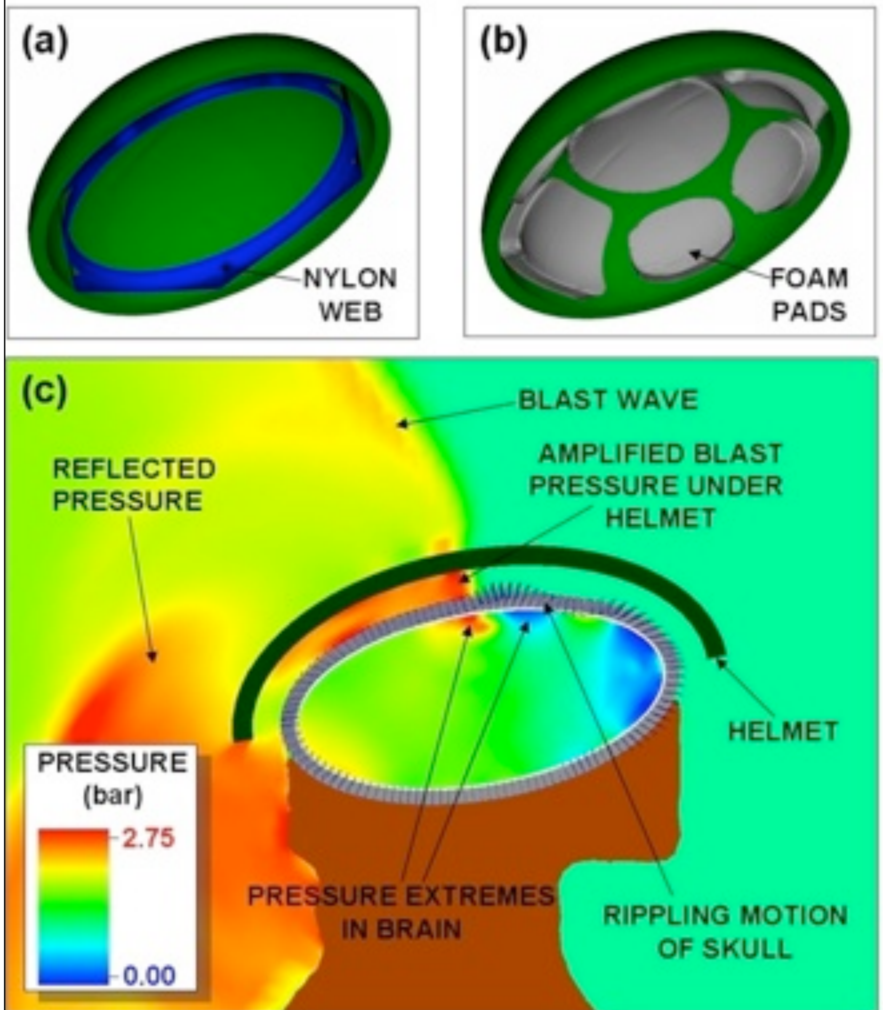
Acceleration of skull during impact
INDUSTRY DAMAGE METRIC



Hydrostatic tension, cavitation
CONTUSIONS/HEMORRHAGING

Moss et al. 09 .

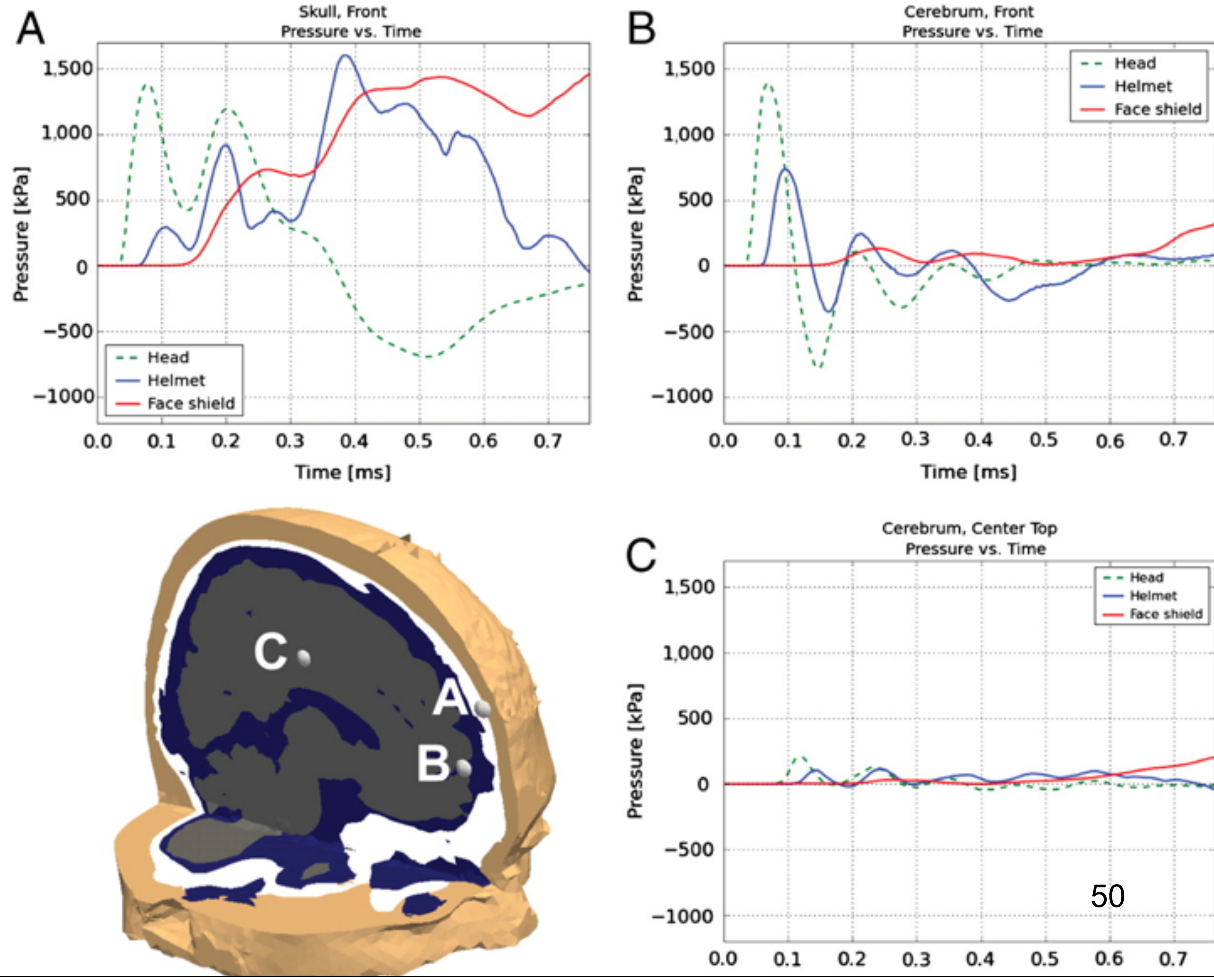
Role of Current Helmets for Blast



- without pads, “underwash” amplfies pressure under helmet: helmet without pads is WORSE than no helmet
- but, with overly stiff pads, head is more strongly coupled to skull and energy is not dissipated in the pads
- need to optimize pad and shell stiffness for both blast + impact

Role of Face Shield (Nyein et al. 2010)

Simulated with unrealistic blast wave, but fairly high fideltiy brain and head model and basic principle probably important: Face shield is probably helpful to deflect blast wave:



Directions/Possibilities

- Improve fidelity of head model (already being done) although some basic lessons learned don’t require an accurate head model: (Helmet protection testing for example)
- Use Impacts recorded from players to get input condtions for computer simulations of head impacts; measure sites in the computational models where forces are maximized and correlate these with DTI images
- can track sites where seemingly repeated injuries are occuring and also correlate those with DTI imaging data
- In short: the simulations become “experiments” that one can use to identify sites of injury and methods of protection
- At present, even without simulations can presently:
  - correlate basic DTI image properties with mechanical metrics
  - correlate crude 2-D information about impact locations with DTI images.

END