BIOMECHANICS: INJURY AND DESIGN CONSIDERATIONS

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1.0 Introduction

Life is full of activities that are engaging, stimulating, and exciting. These activities allow humans to develop physically and mentally, as well as accomplish tasks related to employment, travel and recreation. Activities such as walking, running, cycling, driving, flying, and playing sports are daily occurrences, supplemented by less frequent (but somewhat more exciting) events like roller coaster rides and water slides. While the individual and societal benefits of these activities are numerous (such as increased physical fitness and improved health, decreased health care costs, economic benefit, overall happiness, etc.) there are risks associated with many of these activities. Specifically, these activities pose a hazard of physical injury to individuals, to varying degrees. As a result, design approaches and standards are developed to limit the potential for injury to occur under expected conditions.

When structures such as buildings or bridges are designed, the materials used for construction can be manufactured, processed and tested to accurately characterize their mechanical properties (e.g. stiffness, strength, etc.) and establish the conditions that are required to cause failure. In contrast, designing environments for work, recreation and travel cannot rely on the same type of extensive testing for all the different anatomical structures in the living human body, for obvious moral and ethical reasons. As the properties of an individual's muscle, bone, ligaments and tendons cannot be determined under the vast array of conditions that the human body is exposed to, predictions of the body's response are needed. For this reason, the science of biomechanics, which examines how the body both generates and responds to force, employs a variety of techniques to provide insight into how and why injuries occur. With this approach, biomechanics allows scientists and engineers to design environments, activities and protective equipment to reduce the risk and occurrence of both acute and chronic injuries.

This article highlights the role of biomechanics as a design consideration surrounding a variety of activities and exposures, as well as in less frequent traumatic events. It is hoped that the reader gains a basic understanding of how biomechanics expertise plays a significant and distinct role in the analysis of human-world interaction, in addition to an appreciation of the opportunities and challenges that exist when using predictive methods in the design process.

2.0 Designing Safe Pedestrian Walking Surfaces

Perhaps the most distinct or defining characteristic of humans is our ability to walk upright on two feet, a biomechanical trait known as bipedal gait. However, our ability to walk upright on two feet comes at a cost, due to the inherent instability that is associated with human locomotion. Each step involves lifting one point of contact with the ground (the trailing foot) and rapidly moving that limb ahead of the body in order to support the weight of the body and allow for a subsequent step with the alternate foot. This sequence is repeated, resulting in the characteristic alternating human gait. However, this process requires that the body be supported by a single limb during movement of the trailing foot its location out front. This movement pattern provides an opportunity for a destabilizing event to result in a fall, with significant potential for injury. Fortunately, biomechanics research has provided a great deal of insight into how and why falls occur, which has been applied to improve of safety of our everyday walking surfaces. In general, falls that occur during gait can be divided into two main categories – slips and trips.

2.1 Slips and Falls

Slips occur when there is insufficient friction between the foot of a pedestrian (to be accurate, the material that covers the sole of the foot) and the walking surface. The lack of friction allows the foot to slip relative to the ground, resulting in a loss of balance. While walking, slips generally occur at two points in the gait cycle where the need for friction is the greatest. One of these points occurs during 'toe off', when the trailing foot pushes against the ground to contribute to the forward motion of the body prior to that limb being moved to the front of the body. Slips that occur during this point in the walking cycle result in the trailing foot slipping rearward relative to the body, but do not tend to lead to significant falls. The reason for this is that most of the body's weight has been transferred to the leading foot when the slip initiates, and the body is therefore supported. In addition, slips also initiate at 'heel contact.' This occurs when the swinging foot travels in front of the body and is subsequently brought to the ground. At ground contact, friction is required to slow the forward moving foot. When insufficient friction is available, the swinging foot continues forward and is unable to accept the transfer of body weight. As a result, a fall occurs. Slips at heel contact pose a greater risk of fall initiation than slips at toe off.

To appreciate the biomechanical considerations that govern slip mechanics, it is necessary to understand two key variables in order to prevent their occurrence: (i) how much friction a pedestrian requires, and (ii) how much friction the walking surface can provide. Much research has been dedicated to understanding the amount of friction required for individuals to complete common walking tasks. As an example, Burnfield et al. studied the amount of friction required for healthy individuals and those with various gait deficiencies to walk in a straight line, change direction, ascend and descend stairs.[1] It was found that pedestrians needed an average friction level of less than 0.30 while walking in a straight line, and that this level did not vary significantly across the various levels of disability. The greatest demand for friction existed in stair descent, with average levels of approximately 0.35 to 0.40 needed for the physically disabled participants. These values are consistent with those reported in the literature under similar conditions,[2] and provide a reasonable estimate of the friction requirements under typical walking conditions. However, other factors can play a significant role in the amount of friction required, depending on pedestrian behaviour. As an example, when gait speed is increased, pedestrians apply greater forces to the ground and as a result require greater levels of friction.[3]

When the intended or expected use of a pedestrian facility is established, predictions of pedestrian requirements can be made based on predicted behaviours to reduce the likelihood of a slip occurring. In contrast, when pedestrians behave in a manner that is not consistent with the design of a walking surface, the risk of a slip may be significant. This is a critical consideration, as it is well known that pedestrians can adjust their walking behaviour in response to the known properties of a walking surface.[4] For example, pedestrians can alter their posture (i.e. degree of knee and hip bend), their muscular activity (which affects joint stiffness) and reduce the force with which they strike the ground in order to safely navigate surfaces with obvious low levels of friction (e.g. snow and ice covered side walk) if the conditions are anticipated.

With an understanding of the biomechanics relevant to slip events, walking surfaces can be assessed to determine whether they provide adequate levels of friction to ensure pedestrian

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[1] Burnfield, J.M., et al. 2005. Comparison of Utilized Coefficient of Friction During Different Walking Tasks in Persons With and Without a Disability. *Gait & Posture.* 22: 82-88.
[2] Redfern, M.S., et al. 2001. Biomechanics of Slips. *Ergonomics.* 44: 1138-1166.
[3] Fino, P., and Lockhart, T. 2014. Required Coefficient of Friction During Turning at Self-Selected Slow, Normal, and Fast Walking Speeds. *Journal of Biomechanics.* http://dx.doi.org/10.1016/j.jbiomech.2014.01.032
[4] Heiden, T.L., et al. 2006. Adaptations to Normal Human Gait on Potentially Slippery Surfaces: The Effects of Awareness and Prior Slip Experience. *Gait & Posture.* 24: 237-246.

safety under expected use conditions. However, there is considerable debate within the biomechanics research community regarding the tools and techniques that are employed to measure friction levels, as well as the determination of appropriately ‘safe’ levels. Measuring the friction provided by in-use walking surfaces (either before or after a loss has occurred) inherently requires the use of portable slip testing devices which vary in application, design, and use. Generally, these devices measure the resistance to motion between a rubber test pad and the surface of interest. Some devices measure the ‘static’ level of friction or the amount of force required to move an object divided by the objects weight, while others measure the ‘dynamic’ level of friction, or the amount of force required to maintain the constant motion of an object divided by the objects weight. Research that has compared the array of devices available for such in-use measurements found limited agreement between the devices, and also that most were unable to properly rank surfaces based on ‘slipperiness’, a measure of the number of slips that occurred while pedestrians walked over the surfaces.[5] Some of the available devices require significant user input and can be influenced by experience and training, whereas some are automated. Until recently, devices were not paired with standards relating their measurements of friction to acceptable or sub-standard levels, with some prior standards being rescinded.[6] In 2009, the ANSI/NFSI[7] provided a series of values for wet static testing to be related to measurements obtained with a binary output tribometer (a self-propelled testing device), with values greater than 0.60 being associated with a ‘high traction’ surface. However, historically, a static value of 0.50 has been accepted as the level defining a slip resistant surface; more recently a value of 0.45 has been proposed as slip resistant under wet dynamic test conditions.[8] To date, there has been no consensus on a single definition of slip resistance. Regardless of the test method and/or the definition applied, tests meeting these values would indicate that the walking surface would be slip resistant for walking in a straight line, as the friction level required by pedestrians under these conditions is generally less than 0.30 (as indicated above). If the pedestrian is not walking in a straight line it is necessary to understand the biomechanical

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[5] Powers, C.M., et al. 2010. Validation of Walkway Tribometers: Establishing a Reference Standard. *Journal of Forensic Sciences.* 55: 366-370.

[6] ASTM International F1679-04, Standard Test Method for Using a Variable Incidence Tribometer (VIT).

[7] American National Standards Institute, Inc./The National Floor Safety Institute: B101.1 Test Method for Measuring Wet SCOF of Common Hard-Surface Floor Materials.

[8] Dr. Jens Sebald. System-oriented Conception for the Testing and Assessment of the Slip Resistance of Safety, Protective, and Occupational Footwear. South Lake, Texas, 2008.

demands of the activity with respect to the properties of the walking surface in order to determine the level of risk.

Of course, these field based methodologies do not replicate all the conditions of an incident slip and fall. While it is possible to replicate conditions such as a wet floor, it is often not possible to test the incident footwear (in a non-destructive manner), or to mimic known conditions such as barefoot walking. Further, alterations in the maintenance and cleaning of the incident walking surface, or the application of commercially available anti-slip chemicals between the time of a fall and the time of examination can greatly influence the assessment. Knowledge of these limitations is necessary to allow for proper interpretation of slip resistance measurements.

2.2 Trips and Falls

The mechanics of a trip are distinct from those of a slip, and correspondingly, the hazards that can induce a trip are different. A trip occurs when the swinging foot is impeded in its path from behind the body to the front of the body during a step. As the swinging foot accepts the body weight at the completion of a step, it is relied upon to control the forward motion of the body. If it does not arrive in front of the body as expected, the forward momentum of the body will continue in an uncontrolled manner. As a result, individuals who experience a trip tend to fall forward.[9,10]

Knowing that trips occur as the result of the swinging foot being caught during its forward motion allows for the design of pedestrian walking surfaces that are unlikely to induce such an event. Research has shown that, on average, pedestrians clear the ground with their swinging foot by approximately 1.3 centimeters (0.5 inches).[11] Importantly, the variance of this measurement (expressed as a standard deviation) is roughly 0.62 centimeters (0.25 inches). Stated differently, while the average toe clearance during level walking is 1.3 centimeters, most people will clear the ground by 0.68 to 1.92 centimeters. Specifically, assuming toe clearance is normally distributed among the population (a statistical model common in biology), roughly

[9] Smeesters, C,. et al. 2001. Disturbance Type and Gait Speed Affect Fall Direction and Impact Location. *Journal of Biomechanics.* 34: 309-317.

[10] Troy, K.L., and Grabiner, M.D. 2007. Asymmetrical Ground Impact of the Hands After a Trip-Induced Fall: Experimental Kinematics and Kinetics. *Clinical Biomechanics.* 22: 1088-1095.

[11] Winter, D.A. 1991. Biomechanics and Motor Control of Human Gait. 2nd Edition. Waterloo Biomechanics, Waterloo, Ontario.

68% of the population falls within this range. Using the same statistical assumptions, more than 80% of the population would clear the ground by more than 0.68 centimeters. As a result, walking surfaces need not be perfectly planar to prevent trips from occurring (a standard that would be practically unattainable).

This knowledge is reflected in current standards for the design of safe pedestrian walking surfaces. As an example, the American Society for Testing and Materials (ASTM) F1637, Standard Practices for Safe Walking surfaces allows for height differentials in a walking surface to reach 0.6 centimeters without any treatment, reflecting the known abilities of pedestrians. For height differentials within the range of 0.6 to 1.2 centimetres, a bevelled transition (threshold) is required, reflecting the known biomechanics of human gait, which indicates that a significant portion of the population will clear the ground by this height.

However, despite the significant amount of research that has been conducted on this topic, the known toe clearance ranges are not reflected in all available standards. As an example, in 2010 the Ontario Minimum Maintenance Standards were expanded to consider sidewalks. It was written "*A surface continuity on a sidewalk is deemed to be in a state of repair if it is less than or equal to two centimetres.*"[12] A height differential of 2 centimetres (0.79 inches) is much greater than the expected toe clearance of most pedestrians. As a result, it is predictable based on the biomechanics of human gait that pedestrians will trip on sidewalks that are not considered to be in need of maintenance.

3.0 Biomechanics and the Design of Motor Vehicles

The personal and societal costs associated with injuries sustained in motor vehicle collisions are well known, and do not need to be repeated in this article. Further, the causes of motor vehicle collisions (driver behaviour, weather and road conditions, mechanical failures, etc.) are beyond the scope of biomechanics. However, biomechanics is employed in vehicle design across the immense array of collision circumstances experienced by vehicle occupants, including both low and high speed impacts.

[12] O.Reg. 47/13, s. 16(4).

3.1 Design Features for Low-Speed Collisions

Despite the lack of property damage that is often associated with low-speed rear-end motor vehicle collisions, significant injury complaints have been associated with their occurrence. Scientists have been attempting to understand how and when injuries can be sustained in these events for decades. The difficulty faced by the biomechanics research community is an ethical one, in that studies cannot be designed to expose living people to ever increasing levels of collision severity until an injury is documented. For this reason, studies are often conducted at severity levels considered to be safe, and study participants are asked to report any complaints. Generally, human volunteer studies have found limited complaints for collision severities less than 8 km/h, and in all cases symptoms have been transient for severities less than 14 km/h.[13,14,15,16] The musculoskeletal complaints tend to involve reports of pain and stiffness in the neck and shoulders, as well as headaches. Studies that have been conducted to characterize the mechanism of injury, employing medical imaging in conjunction with simulated rear-end collisions, have revealed the vertebral level changes in spine posture that may induce injury, and have discredited the historical depiction of 'whiplash' occurring due to a significant extension of the neck. What has been found is that the base of the neck is initially driven upward in a rear-end collision, as the thoracic spine (midback) is loaded by the seatback. As a result, the lower levels of the spine are initially forced into extension while the upper spine remains flexed, resulting in the formation of an 'S-shaped' curvature of the neck.[17,18,19] These insights into the mechanics of injury in low-speed rear-end collisions have led to two significant design developments in motor vehicle safety: (i) active head restraints and (ii) energy absorbing seatbacks.

Active head restraints are designed to deploy in response to rear-end motor vehicle collisions, moving forward and upward in response to occupant loading of the seatback. The

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13 Brault, J.R., et al. 1998. Clinical Response of Human Subjects to Rear-end Automobile Collisions. *Archives of Physical Medicine and Rehabilitation.* 79: 72-80.

14 Castro, W.H.M., et al. 1997. Do "whiplash injuries" Occur in Low-Speed Rear Impacts? European Spine Journal. 6: 366-375.

15 Davidsson, J., et al. 2001. Human Volunteer Kinematics in Rear-end Sled Collisions. *Traffic Injury Prevention.* 2: 319-333.

16 Bailey, M.N., et al. SAE Paper 950352. Data and Methods for Estimating the Severity of Minor Impacts.

17 Kaneoka, K., et al. 1999. Motion Analysis of Cervical Vertebrae During Whiplash Loading. *Spine.* 24: 763-770.

18 Ono, K., et al. SAE Paper 973340. Cervical Injury Mechanism Based on Analysis of Human Cervical Vertebral Motion and Head-Neck-Torso Kinematics During Low Speed Rear Impacts.

19 Panjabi, M.M., et al. 1998. Mechanism of Whiplash Injury. *Clinical Biomechanics.* 13: 239-249.

theory is that the active head restraint, by moving towards the head in response to a rear-end collision, minimizes the movement of the head (and correspondingly the neck) reducing the risk of injury. However, the effectiveness of active head restraints has not been definitively supported by the available research. Epidemiologic data has shown a significant reduction in the occurrence of injury with the use of active head restraints,[20] while biomechanical testing of cadavers has shown that the reductions in motion achieved with these design interventions are not sufficient to prevent injury in all levels of the cervical spine.[21]

Energy absorbing seatbacks have also been incorporated into vehicle design to prevent neck injuries in rear-end collisions. These seatbacks are designed to yield (deform) to absorb energy and limit the accelerations that are applied to the occupant. As with active head restraints, the available epidemiologic data supports a reduction in the occurrence of neck injury with the incorporation of such seats in vehicle design; however, biomechanical data indicates that injurious neck motions may still be experienced.[22,23] Confounding the difficulty in designing energy absorbing seatbacks is the need to ensure that occupants are restrained in more severe rear-end collisions, as vehicle designers must ensure that the seatback will not yield excessively and allow vehicle occupants to experience partial or complete ejection if the vehicle is involved in a more severe crash.

3.2 Biomechanics in Understanding Severe Injuries in Motor Vehicle Collisions

In addition to the biomechanical considerations incorporated into the design of motor vehicles to prevent neck injuries in low-speed collisions, there is a need to understand the mechanics of more severe injuries in higher speed collisions. Fortunately, motor vehicles are subjected to controlled crash tests in an effort to ensure occupants will not be exposed to excessive loads under 'typical' collision scenarios. Crash tests involve full width frontal tests, offset frontal tests, and side impacts. For obvious reasons, human volunteers cannot be used in these tests. As a result, anthropomorphic test devices (ATDs or 'dummies') are employed. The

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[20] Viano, D.C., and Olsen, S. 2001. The Effectiveness of Active Head Restraint in Preventing Whiplash. *The Journal of Trauma Injury, Infection, and Critical Care.* 51: 959-969.

[21] Ivancic, P.C., et al. 2010. Effect of Active Head Restraint on Residual Neck Instability due to Rear Impact. *Spine.* 35: 2071-2078.

[22] Ivancic, P.C. Does Knowledge of Seat Design and Whiplash Injury Mechanisms Translate to Understanding Outcomes. *Spine.* 36: S187-S193.

[23] Xiao, M., and Ivancic, P.C. 2010. WHIPS Seat and Occupant Motions During Simulated Rear Crashes. *Traffic Injury Prevention*. 11:514-521.

following discussion will address the design concepts of these dummies and the necessary considerations in relating the measurements obtained from the dummies to the risk of injury.

Due to our inability to subject human volunteers to injury producing events, the use of mechanical analogs for the human body is common place in biomechanics. Perhaps the most commonly used test dummy in motor vehicle research is the Hybrid III test dummy. This dummy was originally developed by General Motors to represent the physical properties and response of an average (50th percentile) male to frontal collisions.[24] The dummy is instrumented to measure head accelerations, chest deflections, and femur (long bone in the thigh) forces in response to collisions. The Hybrid dummy has also been scaled to represent a large male (95th percentile), a small female (5th percentile), and 3, 6, and 10 year old children. In establishing the fidelity of these dummies, researchers have compared the response of these dummies to cadavers (post mortem human surrogates or 'PMHS') under a variety of test conditions.[25,26] However, the Hybrid III dummies are not designed to be biofidelic under all crash conditions, and as a result an array of crash test dummies have been developed with various mechanical differences. As an example, for side impact testing there is a European side impact dummy (EuroSID) and a Worldwide harmonized Side Impact Dummy (WorldSID). Further, there exists rear impact dummies (RID's) specifically designed to allow for more accurate representation of local neck motions, such as the BioRID. Each of these test dummies has been designed to replicate the motions experienced by humans under a specific set of collision conditions, providing an opportunity to document the loads experienced throughout the body.

The loads measured by the dummies under the controlled crash conditions are compared to established predicted injury tolerances for each measured region of the body. These injury values, referred to as Injury Assessment Reference Values (IARVs) are based on cadaveric testing performed to determine how much load the body can tolerate. There are several issues that complicate this process and the interpretation of risk values. First, and this is a fortunate reflection on society, much of the available cadaveric data is obtained from elderly cadavers. It

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[24] For a historical review, the interested reader is directed to Nahum, A.M., and Melvin, J.W. (Ed). 2002. Accidental Injury: Biomechanics and Prevention. 2nd Edition. Springer-Verlag, New York, USA.

[25] Beeman, S.M., et al. 2013. Kinetic and Kinematic Responses of Post Mortem Human Surrogates and the Hybrid III ATD in High-speed Frontal Sled Tests. *Accident Analysis & Prevention.* 55: 34-47.

[26] Vezin, P., et al. 2002. Comparison of Hybrid III, Thor-α, and PMHS Response in Frontal Sled Tests. *Stapp Car Crash Journal*. 46: 1-26.

is well known that the strength of tissues changes with age, and therefore care must be taken when relating the predicted tolerance of a tissue to the loads experienced by a younger individual. Second, the IARVs are related to a percentage risk, not an absolute injury level. For example, the allowable femur load for a 50th percentile male in a frontal collision is 10 kiloNewtons (kN) according to FMVSS 208. However, this does not mean that loads less than 10 kN will not cause injury. At this force level, it is predicted that the risk of a moderate or greater injury for an average male is 35%.[27] Predicting the risk of injury using IARVs also assumes that the threshold for injury for the injured area of the body is known. As an example, the risk of head injury in crash testing is based upon the Head Injury Criteria (HIC) which considers both the magnitude of the head acceleration experienced as well as the duration of the exposure. This method of estimating injury tolerance was based largely on historical head impact data, wherein the focus was on injury to the rigid structures of the head (the bones of the skull) and the occurrence of severe injury, with associated inferences of brain injury. With the development of restraint systems, rigid impacts to the head are becoming less frequent, while concurrently our understanding of mild traumatic brain injuries (e.g. concussion) is increasing. As a result, employing an injury metric such as the HIC to quantify the risk of a head injury may lead to the correct conclusion that a serious head injury would not be expected, but may miss the potential for another real injury, such as a mild traumatic brain injury.

Finally, the use of IARVs assumes that a single threshold value can be applied to a range of individuals. Generally, IARVs are available for the average male, small female, 6 year old child, 3 year old child, and 12 month old. However, these estimates of IARVs are based on dummies designed to reflect the 'average' proportions of individuals of these sizes. Unfortunately, individuals rarely are 'average' across all dimensions (e.g. height and weight) and therefore may not be geometrically simulated using the available crash test dummies. Acknowledging these limitations, researchers have proposed scaling techniques to adjust measured loads and IARVs for occupant size, with the research ongoing.[28]

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[27] Kleinberger, M., et al. 1998. Development of Improved Injury Criteria for the Assessment of Advanced Automotive Restraint Systems. www.nhtsa.dot.gov.

[28] Yoganandan, N., et al. 2014. Normalizing and Scaling of Data to Derive Human Response Corridors from Impact Tests. Journal of Biomechanics. http://dx.doi.org/10.1016/j.jbiomech.2014.03.010.

Despite the above considerations in relating the forces as measured with an ATD to the risk of injury for a particular individual, it is clear that an ever increasing understanding of injury biomechanics is improving the safety of motor vehicles.

4.0 Biomechanics in the Design of Safe Workplaces

Since a great deal of an individual's time is spent completing the tasks associated with their employment, it is necessary to understand when and how injuries occur in the workplace to ensure that employment tasks are designed to minimize such risks. Two major sources of work-related injuries are slip and trip events, the biomechanics of which has been addressed above. In addition, work-related injuries can result from an obvious trauma such as a fall from height or interaction with heavy machinery, wherein the source of the injury is relatively straightforward. Finally, injuries are also sustained in the workplace during manual materials handling tasks (lifting and carrying) due to both over-exertion and fatigue mechanisms, where the cause of the injury is less obvious. An understanding of the biomechanics of injury related to over-exertion and fatigue allows for the design of safer workplaces and a reduction in the occurrence of preventable injuries.

4.1 Over-Exertion Injury

Over-exertion injuries are similar to traumatic injuries in that they occur when the forces of a task exceed the tolerance of the body during a single event. Such an injury may occur if an individual attempts to complete a heavy lift that is beyond their capability. These injuries can be prevented if the tolerance of the body and the loads being experienced can be reasonably estimated.

In manual materials handling tasks, it is often the low back that is at risk of injury due to the significant forces generated by the spinal musculature. As a result, significant efforts have been devoted to estimating individual tolerance to loading, based on factors such as age, gender and spinal level (the region of the spine). Typically, this research involves exposing a sample of isolated spines to controlled doses of compressive force (vertical loading, as occurs in lifting) until failure occurs (e.g. endplate fracture), which enables the capacity of the specimen to be compared across individual characteristics. By studying the spine in this manner, researchers have developed predictive equations to allow for individualized estimates of spine tolerance.[29,30] Interestingly, researchers have attempted to further improve their ability to estimate an individual's tolerance to load by incorporating direct measurements of the bone mineralization. This approach uses medical imaging to measure the bone mineral content of the vertebrae, and then relates this quantity to the strength of the spine. However, it has been shown that the inclusion of this level of detail has not led to significant improvements in the ability to estimate strength above more simple measures of geometry.[31]

Regardless of how the estimate of strength is generated, a comparison is made to the internal loads experienced while completing the task. Clearly it is not possible to simply measure the forces within an individual's spine while they complete their work. For this reason, biomechanical models are constructed to estimate the internal loading. These models vary in complexity, and can include the forces in the hands, the posture of the individual, as well as the muscular activation required to complete the task. The more simple models are suitable for use in the field (i.e. in the workplace); while the more sophisticated approaches can only be employed in the laboratory due to the need for extensive measurement. The simplest approach is to assume that all of the muscles can be represented by a single entity, known as the 'single muscle equivalent,' which allows for incorporation of the muscle forces in estimates of loading. This is an assumption of convenience and lacks biological fidelity; however, it can provide reasonable estimates of loading under certain conditions, such as lifts performed slowly with limited contribution from opposing muscle groups, such as the abdominal muscles. This approach has been integrated into well- developed publically available software used in designing safe jobs.[32] At the other end of the spectrum are models that better reflect the anatomy of the spine, but require a great deal of input information.[33] Research has found significant

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[29] Genaidy, A.M., et al. 1993. Spinal Compression Tolerance Limits for the Design of Manual Material Handling Operations in the Workplace. *Ergonomics.* 36: 415-434.
[30] Davis, K.G., and Parnianpour, M. 2003. Subject-Specific Compressive Tolerance Estimates. *Technology and Health Care.* 11: 183-193.
[31] Parkinson, R.J., et al. 2005. Estimating the Compressive Strength of the Porcine Cervical Spine. An Examination of the Utility of DXA. *Spine.* 30: E492-E498.
[32] 3D Static Strength Prediction Program, Version 6.0.6. The University of Michigan Center for Ergonomics.
[33] McGill, S.M., and Norman, R.W. 1986. The Volvo Award for 1986: Partitioning of the L4/L5 Dynamic Moment into Disc, Ligamentous and Muscular Components During Lifting. *Spine.* 11: 666-678.

differences in employing various models to analyze a given task,[34,35] and it is incumbent on the expert to determine the modelling approach appropriate to the task.

Once the loads on the spine have been estimated according to the task demands and the individual, they can be compared to the estimated tolerance of the worker. If the tolerance of the worker is greater than the loads on the body, there is no risk of injury. If the loads are predicted to exceed the tolerance of the body, the job should be redesigned to prevent injury from occurring. However, as described above it is not trivial to estimate the internal loads and tolerances, so simplified approaches have been widely adopted. For example, the U.S. based National Institute of Occupational Health and Safety (NIOSH) has developed an equation to estimate the 'recommended weight' of a lift based only on the physical dimensions of the task (i.e. how far the object is from the body, how high it is to be lifted, if the object is symmetrical, etc.).[36] Using this approach, the maximum weight to be lifted by any employee is 23 kg. This mass has been estimated to result in a spine compressive tolerance of 3400 N, which was estimated to protect the majority of workers. However, this approach is often misapplied and misinterpreted as a means to estimate an individual's risk of injury. Exceeding the NIOSH recommended weight limit for a lifting exposure does not imply that an individual would have experienced loading in excess of their tolerance, as the NIOSH approach does not incorporate individualized estimates of spine loading and has been shown to underestimate the tolerable loads.[37]

One fascinating area in biomechanical assessment involves consideration of the role of the individual in modifying the work-related loads on the body. While theoretically the loads on the body can be solely a function of the task demands (the weight of the object to be lifted, the height of the lift, the position of the body, etc.), in reality the individual performing the task can alter the loads on their body significantly. It has been shown that factors such as gender, stress

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[34] Parkinson, R.J., and Callaghan, J.P. 2009. The Use of Artificial Neural Networks to Reduce Data Collection Demands in Determining Spine Loading: A Laboratory Based Analysis. *Computer Methods in Biomechanics and Biomedical Engineering.* 12: 511-522.

[35] Parkinson, R.J., Bezaire, M., and Callaghan, J.P. 2011. A Comparison of Low Back Kinetic Estimates Obtained through Posture Matching, Rigid Link Modelling and an EMG-assisted Model. *Applied Ergonomics.* 42: 644-651.

[36] Waters, T.R., et al. 1993. Revised NIOSH Equation for the Design and Evaluation of Manual Lifting Tasks. *Ergonomics.* 36: 749-776.

[37] Potvin, J.R. 2014. Comparing the Revised NIOSH Lifting Equation to the Psychophysical, Biomechanical and Physiological Criteria Used in its Development. *International Journal of Industrial Ergonomics.* In Press.

and even personality type can all alter the loads within the body despite the external task demands being constant.[38]

4.2 Fatigue Injury

The above discussion focussed on injuries that occur when the forces involved are large. With the ability to estimate forces and knowledge regarding injury mechanisms, workplaces have implemented ergonomics programs to reduce or eliminate the need for workers to experience such strenuous tasks, in an effort to prevent over-exertion injuries. This is reflected in the proliferation of workplace tools and mechanisation observed in modern workplaces. However, musculoskeletal injuries still occur, despite best efforts that have been applied in prevention. One mechanism of injury that is often overlooked is fatigue. With fatigue injury, the tissues of the body are degrading over time (i.e. accumulation of damage), losing their ability to withstand loading. If the tissue breakdown is not repaired through adequate rest and recovery, this degradation can continue until the individual performs a common, low load task (most often one they have completed daily for years) with a significant injury outcome. In these circumstances, it is not the specific task that caused the injury (i.e. the straw that broke the camel's back) but the lifetime loading history of the individual.

One of the most common areas of the body for this injury mechanism to occur is in the intervertebral discs of the spine. These discs act to transmit load between the bony vertebrae while allowing the necessary motions to accomplish daily tasks. The discs are comprised of fibrous outer layers (known as the annulus) and a viscous internal centre (known as the nucleus). With repetitive loading, the nucleus can work its way through the layers of the annulus away from its initial position within the centre of the disc to the outer layers. This can lead to disc bulging, changes in disc height, and if sufficient, intervertebral disc herniation. Researchers have been trying to determine the mechanism by which this injury occurs for decades. Initially, it was thought that these disc injuries were the result of rapid overload of the spine (i.e. a trauma). However, researchers found that the spine would need to be excessively flexed (bent forward as in touching one's toes) for this to occur,[39] as under physiological loading it is the

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[38] Marras, W.S., et al. 2000. The Influence of Psychosocial Stress, Gender, and Personality on Mechanical Loading of the Lumbar Spine. *Spine.* 25: 3045-3054.

[39] Adams, M.A., and Hutton, W.C. 1982. Prolapsed Intervertebral Disc. A Hyperflexion Injury. *Spine.* 7: 184-191.

vertebrae that fracture first.[40] This led to the examination of fatigue processes as the likely explanation for disc herniation, and it was found that repetitive bending of the spine, without the need for significant compressive loading, resulted in disc herniations.[41] Interestingly, this process can go on undetected within the spine, as the inner region of the intervertebral disc does not contain pain sensing nerve fibres. This is reflected in the medical imaging of asymptomatic individuals, which has found intervertebral disc injury despite the absence of any pain or clinical symptoms.[42,43]

Another area of the body prone to fatigue injuries is the rotator cuff, a set of four small muscles that act to position the head of the humerus (long bone in the upper arm) within the socket of the shoulder blade (scapula). In particular, the supraspinatus muscle is at risk with frequent, repetitive work. This muscle runs along the upper portion of the shoulder blade and below a bony prominence, known as the acromion process. The process essentially forms the roof of a tunnel through which the muscle runs. When work is performed with the hands at or near shoulder level this muscle is activated and can become compressed within the bony tunnel, resulting in reduced blood flow, loss of nutrient exchange, cellular breakdown and subsequent muscular tearing. As with the intervertebral discs of the spine, this process can be undetected by the individual and tearing within the rotator cuff has been observed in the asymptomatic population.[44,45]

Armed with the knowledge of the mechanics of fatigue injuries, workplace design strategies have been implemented in an attempt to prevent the occurrence of prolonged repetitive loading of individual tissues. Much of this work has focussed on administrative processes, such as job rotation, wherein a worker does not complete the same task for an entire shift but instead rotates between job stations designed to load different areas of the body selectively. However,

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[40] Parkinson, R.J., and Callaghan, J.P. 2009. The Role of Dynamic Flexion in Spine Injury is Altered b Increasing Dynamic Load Magnitude. *Clinical Biomechanics.* 24: 148-154.

[41] Callaghan, J.P., and McGill, S.M. 2001. Intervertebral Disc Herniation: Studies on a Porcine Model Exposed to Highly Repetitive Flexion/Extension Motion with Compressive Force. *Clinical Biomechanics.* 16: 28-37.

[42] Jensen, M.C., et al. 1994. Magnetic Resonance Imaging of the Lumbar Spine in People without Back Pain. *The New England Journal of Medicine*. 331: 69-73.

[43] Matsumoto, M., et al. 1998. MRI of Cervical Intervertebral Discs in Asymptomatic Subjects. *The Journal of Bone and Joint Surgery [British]*. 80-B: 19-24.

[44] Moosmayer, S., et al. 2009. Prevalence and Characteristics of Asymptomatic Tears of the Rotator Cuff. An Ultrasonographic and Clinical Study. *The Journal of Bone and Joint Surgery [British]*. 91-B: 196-200.

[45] Miniaci, A., et al. 1995. Magnetic Resonance Imaging Evaluation of the Rotator Cuff Tendons in the Asymptomatic Shoulder. *The American Journal of Sports Medicine.* 23: 142-145.

immense opportunities remain to better understand the complex interactions between mechanical exposures, rest and recovery and an individual's risk of fatigue related injury.

5.0 Summary

The above discussion was intended to highlight several areas where an understanding of the mechanics of the human body can facilitate the development of environments, products and activities that minimize the potential for injury. The science of biomechanics, while widely applicable, is tasked with overcoming a significant difficulty – the inability to design, manufacture and test all of the tissues that make up the body. Instead, biomechanics relies on tools such as sub-threshold human testing, Anthropomorphic Test Devices, cadaver testing and numeric modelling to predict when and how injuries will occur. These tools have led to improved awareness of design standards and approaches that reflect human needs. The reality is that humans will continue to undertake activities that pose a risk of injury – such as locomotion, operation of motor vehicles, work, and recreation – and therefore the design of such tasks must address the known biomechanical principles that govern injury. It would be a significant loss to society if humans stopped participating in the activities that generate health and happiness because of the risks associated with everyday activities.