MUSCULOSKELETAL LOADING AND IMPLICATIONS FOR INJURY

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There are numerous variables used to assess musculoskeletal loading during human movement. This presentation will examine ground reaction forces, segment accelerations, joint contact forces and internal bone stresses and strains. I will cover implications for injury assessment, subtleties of interpretation, benefits and drawbacks of each these methods.

INTRODUCTION:

Sprained ligaments, torn connective tissue, contusions, fractured bone and strained muscles are the result of excessive loads placed on biological tissues. These loads generally cannot be measure directly so we are left with a search for surrogate variables that we can measure. Results are often less than satisfactory because, by the nature of being a surrogate, these variables do not behave in the same way that direct measurement of the injury causing loads behave. For instance, we often use the vertical ground reaction force as a measure of the overall loading on the body during running activities. However, the only tissue that actually receives these loads is the bottom of the foot. The effects of the vertical ground reaction force (2.0-2.5 body weights (BW)) are overwhelmed by muscular forces (Achilles tendon force: 6.1-8.2 BW, Scott and Winter, 1992) further up the skeletal system. Thus, reducing the loads caused by the vertical ground reaction force may be detrimental if it results in increased muscle loads. In this paper I will explore some of the benefits and limitations of using surrogate measures to estimate the potential for injury due to overloading of tissue.

GROUND REACTION FORCES:

Ground reaction forces are easily obtained and very reproducible. During activities that involve running there is generally an impact peak that occurs within the first 50 ms after contact with the ground. This peak is due to the deceleration of the leg or a portion of the leg. The mass being decelerated during the impact is called the effective mass. A second peak occurs during midstance as a result of the deceleration of the rest of the body. This peak is often called the active peak because it is more directly under muscular control. In Figure 1 the vertical ground reaction (VGRF) curve has been decomposed into impact and active components (Derrick *et. al*, 2005). The magnitude of the VGRF impact peak is often used as the best measure of detrimental loading to body. This measure has been criticized because it is relatively small (see Introduction) and insensitive to alterations. For instance, replacing a hard shoe with a soft one will have two effects on the impact curve. It will reduce and delay the peak value. When the active and impact curves are temporally summed these two effects can cancel.

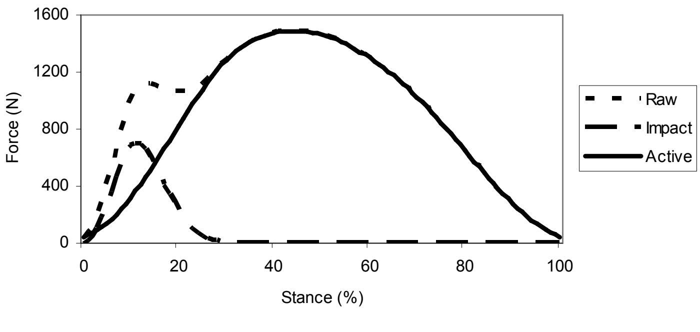

Figure 1: Decomposition of the vertical ground reaction force

ACCELERATIONS:

Part of the issue with ground reaction forces is that they measure both the active and impact forces simultaneously. Accelerometers can be attached to specific locations on the body and therefore reflect the loading of just that portion. Accelerometers attached to the leg will be a more direct measure of the impact without the influence of the active loading. However, the acceleration of the leg is only proportional to the forces causing the acceleration when the effective mass is constant. Effective mass is reduced when a runner makes contact with the ground while using a greater amount of knee flexion (Derrick, 2004). Thus, it is difficult to determine if an increase in peak acceleration is the result of increased force or decreased mass. In a recent study examining the effects of midsole hardness and running surface we found that the harder midsole increased peak acceleration values by about 0.6 g's. However, the softest running surface (sand) caused runners to increase knee flexion and therefore decrease effective mass. Running on the sand had the greatest peak acceleration values of any of the 8 surfaces tested; 0.9 g's greater than running on cement.

ATTENUATION:

Attaching a second accelerometer to the body allows the calculation of attenuation. This is the reduction of the impact as it is transmitted through the skeletal system. An accelerometer attached to the leg will register impacts of 5-10 g's. By the time the shock wave has reached the head, the magnitude has been reduced to 1-2 g's. Typical attenuation values are 74-85%. This method underestimates the attenuation because the head accelerometer contains an active component similar to the vertical ground reaction force curve. An alternative method for calculating the attenuation is in the frequency domain (Shorten and Winslow, 1992). This method separates the high frequency impact component from the lower frequency active component before calculating the impact attenuation (Figure 2).

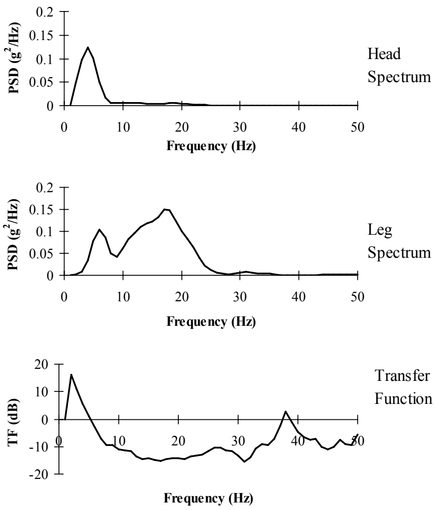

Figure 2: Power spectral densities and the transfer function for running

JOINT CONTACT FORCES:

Modeling the body as a stick figure (rigid body model) has been a popular way to simplify the system enough to estimate the forces that occur within the body. The forces acting on the foot segment (ground reaction forces) are combined with an anthropometric model (mass, center of mass and moments of inertia) and kinematics to estimate the net forces and moments acting at the ankle. The process then moves to the leg segment and then the thigh segment to estimate all of the lower extremity joint moments and joint reaction forces. The joint reaction forces estimated by this process are not the actual forces acting at the joint. As an example, consider the force at the ankle during static standing. This force is simply the weight of the body minus the weight of the foot. However, co-contracting the ankle plantar and dorsi flexors will increase the actual forces at the ankle without increasing the joint reaction forces. The actual forces (joint contact forces) acting at the ankle are calculated by summing the joint reaction forces with the muscle forces.

Estimating the muscle forces is difficult because there are many different muscles and therefore many possible sets of muscle forces that will create the joint moments calculated with the rigid

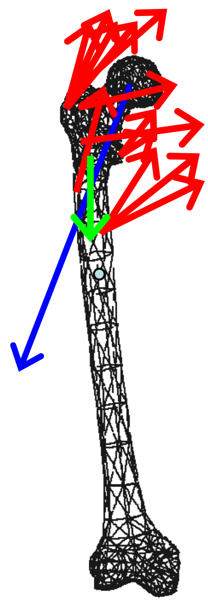

Figure 3: Muscle, contact and gravitational forces acting on the upper half of the femur during running

body model. A cost function is utilized to reduce the solution set. This cost function is a guiding principle that is used to determine which muscles to contract. If we assume that the body chooses the set of muscles that minimizes the total stress squared or stress cubed then we can reduce the possible outcomes to a single set of muscle forces. Optimization procedures have been developed that help find these muscle forces. Once the muscle forces have been summed with the joint reaction forces we have an estimate of the actual loading that takes place at the joints. These forces can be useful in assessment of injury potential at the joint but there are other areas of the lower extremity that can be injured. Further modeling must be utilized to examine the forces within bone.

INTERNAL BONE FORCES:
In order to calculate the forces and moments within the bone the joint contact forces, muscle forces and the weight vector (Figure 3) must be known. The internal bone forces (Figure 4) indicate the compressive and shear forces acting at the centroid of the bone. The internal bone moments indicate the tendency of the bone to bend or torque. Both the forces and the moments contribute to the loading on the periphery of the bone. Bending moments will create compressive loading on the concave side of the bend and tensile loading on the convex side.

BONE STRESSES AND STRAINS:
Strains can be measured directly via strain gauges attached to the bone or to pins that are drilled into the bone. This works best on cadavers but some labs have successfully measured strains in live humans performing walking, running and landing tasks. Stress estimates can be obtained by combining forces and moments into a single value assessment of the loading environment if the bone geometry is known. If the material properties of the bone are known then strains can be calculated from the stress estimates. Figure 5 shows that the stresses on the superior aspect of the neck of the femur increase as the gluteus medius muscle forces are decreased from 100% to 0% of the actual values. The gluteus medius muscle counters the bending of the neck of the femur that is caused by the weight of the torso. This is a possible mechanism for the high incidence of femoral neck stress fractures in runners.

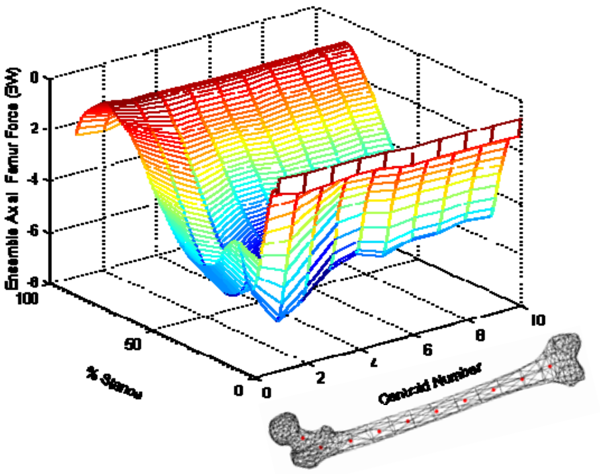

Figure 4: Internal bone forces at specific centroid locations

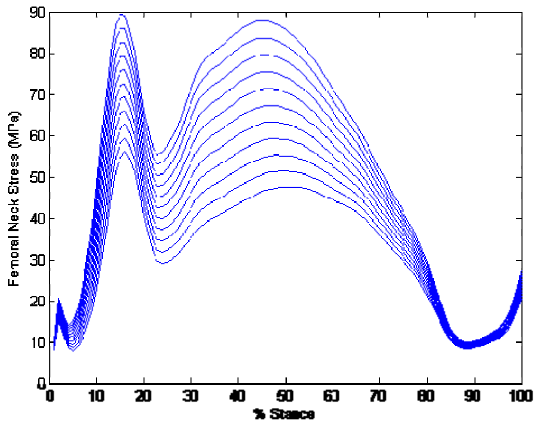

Figure 5: Femoral neck stresses increase when the gluteus medius muscle force is decreased

DISCUSSION:
Although internal loading of the skeletal system is difficult to estimate, there are some major advantages of this process:

1. Loading can be determined at the sites of injury.

2. All of the components of loading can be accounted for (external forces, muscle forces, bending moments, axial forces, shear forces, etc.).
3. No invasive methods are required.

There are also some notable disadvantages:

1. Lots of modeling, therefore lots of assumptions.
2. Individualized models are difficult to construct.
3. Time consuming.

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