Minisymposium: Biomechanics for the FRCS Orth Exam

(iv) Basic biomechanics of human joints: Hips, knees and the spine

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**Summary**

The paper provides a basic introduction to the biomechanics of the hip, knee and spine with respect to the healthy joint and following joint replacement. The content is aimed specifically at persons with a medical background to introduce them to the concepts of forces and moments in application to the human body.

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Basic principles of mechanics (forces and moments)

Forces and moments (torque) can be described by referring to the child’s seesaw in Fig. 1. Standing stationary a child will exert a force onto the ground which is equal to the product of the child’s mass and the acceleration due to gravity (9.81 m/s\(^2\)). Therefore, a child of 30 kg in mass would exert a downward force of 294.3 N to the ground (their weight). Note that if the child jumped onto the ground the acceleration term in the equation (force = mass \(\times\) acceleration) would increase, as would the force.

A force acting on a body can produce rotation as well as translation. This rotation is caused by a rotational torque, also referred to as a moment. A moment is caused when a force acts at a particular distance from a point of reference. This point of reference may be a fixed axis of rotation as in a ceramic-on-ceramic artificial hip replacement or variable as is the case in the functional spinal unit or knee. The magnitude of the moment is calculated by multiplying the force by the perpendicular distance from the line of action of the force vector to the reference point. The seesaw example below explains the basic principles. If Child A sits on the left end of the seesaw the bar will rotate downward, in the counterclockwise direction. In engineering terms, counterclockwise rotation is termed a positive moment, or positive torque. The magnitude of this torque would be equal to the product of the weight of Child A (294.3 N) multiplied by the distance (\(a\)) the child is sitting from the centre of rotation of the beam (2 m), in this case the magnitude of the moment would be equal to 588.6 Nm (294.3 N \(\times\) 2 m). The distance itself is also termed the moment or lever arm.

A moment or torque is defined by a magnitude and a direction and is, therefore, a vector. If a heavier child (Child B 60 kg in mass) sits on the other side of the seesaw the beam will then rotate in the clockwise direction. For equilibrium to occur the sum of the moments acting on the bar must be equal to zero. Therefore, for the bar to be
balanced the product of the weight of Child A and distance \( a \) must be equal to the product of the weight of Child B and the distance \( b \). As Child B is twice the weight of Child A this means that Child B must slide up the bar until his moment arm (distance \( b \)) is equal to one-half of the moment arm (distance \( a \)) of Child A for the seesaw to function effectively. The moments would then be said to be equal in magnitude but opposite in direction. To calculate the reaction force at the pivot of the seesaw you can add up all of the forces acting in the vertical direction. This reaction force would then be equal to the combined weight of the two children.

The hip

The bony structures and ligaments of the natural hip create essentially a ball-in-socket joint. This structure limits anterior/posterior, and medial/lateral translation as well as subluxation (dislocation); however, it does not generally limit the range of motion of the hip during normal daily activities. The allowable range of motion is shown in Table 1 when compared to a selection of daily activities. The range of motion of the hip is far greater than what is required for normal activities, such as walking. This means that the surrounding bone and ligaments of the hip joint do not provide any rotational stability to the hip joint during the walking cycle and, therefore, this stability is provided almost entirely by the action of muscle forces. Principles of simple static mechanics can be used to analyse the loading applied within the body. Figure 2 shows a very simple two-dimensional (2D) schematic of the leg at the heel-strike phase of gait. Contact with the ground produces a ground reaction force equal to the proportion of the person's mass transferred to the ground multiplied by the acceleration of this mass (gravitational acceleration+linear acceleration). The ground reaction force at heel-strike can be measured experimentally using a simple force platform and was reported by Bassey et al.\textsuperscript{6} to be in the region of their body weight (BW). As the knee is fully extended at heel-strike the leg can be analysed in a similar manner to the seesaw as shown in Fig. 2 with the hip joint acting as the pivot. The ground reaction force acting at the foot (\( F_{gr} \)) produces a counterclockwise (positive) moment about the hip centre equal to \( F_{gr} \times Gr \). The moment arm \( Gr \) is equal to the length of the leg (\( L \)) multiplied by the sine of the flexion angle (30°); \( Gr \) is, therefore, equal to 0.5\( L \). Thus, the ground reaction force (\( F_{gr} \)) produces a flexion moment of \( BW \times 0.5L \).

The flexion moment is balanced by the extensor muscles including the gluteus maximus and the hamstrings which act to stabilise the hip at heel-strike producing a counter-clockwise negative moment. In this example, for simplicity, the hamstrings muscle alone has been considered. The moment produced by the hamstrings muscle is equal to the magnitude of the hamstrings muscle force (\( F_h \)) multiplied by the distance from the line of action of the muscle to the hip centre. Thus, the moment arm will vary from person to person; however, its value will be approximated in this example to be equal to 0.15\( L \). Therefore, the moment produced by the hamstrings muscle group would be equal to \( F_h \times 0.15L \). For the hip to be stable the ground reaction moment (\( BW \times 0.5L \)) must be approximately equal to the moment produced by the extensor muscles, in this case only the hamstrings. Therefore, the hamstrings muscle force \( F_h \) at gross approximation would be equal to 3.3 times BW.

The vertical reaction force at the hip can be calculated by summing the forces acting in the vertical direction from Fig. 2. This includes the ground reaction force \( F_{gr} \) and the vertical component of the hamstrings muscle force vector \( F_h \cos(30+\theta) \), where \( \theta \) is the angle between the hamstrings muscle force vector and the line of action of the femur. In this case, \( \theta \) is assumed to be equal to 17°. Therefore, for this simple example, the vertical reaction force at the hip is

\[
\text{Vertical Reaction Force} = F_{gr} + F_h \cos(30+\theta)
\]

The table below shows the range of motion (degrees) in the hip compared to daily activities.

<table>
<thead>
<tr>
<th>Allowable\textsuperscript{1}</th>
<th>Walking\textsuperscript{2,3}</th>
<th>Tie Shoe\textsuperscript{4}</th>
<th>Stairs\textsuperscript{5}</th>
</tr>
</thead>
<tbody>
<tr>
<td>Flexion/extension</td>
<td>140/30</td>
<td>30/15</td>
<td>129</td>
</tr>
<tr>
<td>Internal/external rotation</td>
<td>90/90</td>
<td>4/9</td>
<td>18 abd.</td>
</tr>
<tr>
<td>Abduction/adduction</td>
<td>90/30</td>
<td>7/5</td>
<td>13 ext.</td>
</tr>
</tbody>
</table>

\textsuperscript{1}\textsuperscript{1}\textsuperscript{1} Allowable range of motion in the hip compared to daily activities.

\textsuperscript{2} Walking range of motion in the hip.

\textsuperscript{3} Tied shoe range of motion in the hip.

\textsuperscript{4} Stair climb range of motion in the hip.

\textsuperscript{5} Data from Bassey et al.\textsuperscript{6}
equal to BW+3.3BW × 0.68 for a total of 3.25BW. This 2D calculation is very crude; however, it demonstrates the general methods which can be utilised to analyse forces and moments in the body. It also demonstrates the inefficiency of our muscle forces due to the nature of our relatively long slender limbs and the resulting short muscular moment arms.

The vertical hip joint reaction force during walking is shown in Fig. 3. The walking cycle is characterised by two peaks of load at heel-strike and at toe-off which generally range from 3–4 times our body weight\(^7\). Between these two loading peaks the body’s mass (head and torso) is moving smoothly and is not translated vertically a large amount. As such the reaction force at the hip between heel-strike and toe-off is relatively small and in the region of body weight. At toe-off the hip is extended 15 degrees. The quadriceps muscle acts to stabilise the knee whilst the gastrocnemius muscle produces plantarflexion at the ankle. These muscle forces combine to accelerate the body forward producing the second peak of load in the reaction force curve.

**Figure 3** Vertical reaction force at the hip used for simulation of gait in wear simulators.

**Figure 4** Simplified schematic of standing demonstrating the concept of femoral offset.

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**Hip replacement**

As the wear of replacement joints has improved over the past 20 years, correct positioning of the components during hip replacement is arguably the most important factor for the success of modern total hip prostheses. To restore normal function in the hip joint, an important biomechanical consideration in total hip replacement is the femoral offset. Normal function is in itself an arguable quantity as what one patient would consider normal may limit the activities of another.

The femoral offset is the distance from the centre of rotation of the hip joint to the line of action of the femur, as outlined in Fig. 4.\(^8\) To understand the importance of offset consider the example shown below of a person standing on one leg whilst their body remains vertical. The BW of the person will create a clockwise (negative) moment or torque about the centre of the hip. This torque must be balanced by an equal but opposite counterclockwise (positive) moment produced by the abductor muscle force. The moment arm of the abductor muscle is directly related to the magnitude of
the femoral offset. As the offset increases in length the force required by the abductor muscles to balance the BW moment would reduce, thereby increasing the efficiency of the abductors. The reaction force at the hip would also decrease in this case since the sum of the forces acting in the vertical direction would reduce with the smaller abductor muscle force. Increasing the femoral offset may result in increased stress transferred to the femoral component and its fixation due to the larger bending moment produced by a longer neck length, however, as this is combined with a reduced joint load the overall effect on the stress distribution would be more complex with greater bending stress, but, reduced normal stress, a topic which is beyond the scope of this paper.

The knee

The knee joint consists of two articulating joints the tibio-femoral joint and the patello-femoral joint. Unlike the ball-in-socket geometry of the hip the femoral and tibial surfaces of the knee are not a close fit to one another. The variation in geometry allows a wide range of motion to occur which allows us to complete various daily activities. The allowable range of motion in the knee is shown in Table 2 when compared to a selection of daily activities.

The corresponding reaction forces at the knee during walking are shown in Fig. 5. The reaction forces must be considered in parallel to the flexion–extension of the knee (Fig. 6) in order to fully understand their significance towards the characteristic three peaks of load which occur during walking.

At heel-strike contact with the ground produces a flexion moment at the hip, and an extension moment at the knee, both of which are resisted by the hamstrings muscle. At this position, the knee is fully extended and there is a loading peak across the joint upon impact of ~2–4 times BW, primarily due to muscle forces acting to stabilise the knee.9,12 When our heel hits the ground the knee is in its most stable position due to three factors.1,10 The medial/lateral spacing of the femoral condyles is the least when the knee is fully extended. At this position, the condyles tighten against the intercondylar notch (tibial spine) thereby providing bony stability. The radius of curvature of the femoral condyles is

<table>
<thead>
<tr>
<th>Joint Type</th>
<th>Allowable1</th>
<th>Walking8,10</th>
<th>Sitting11</th>
<th>Stairs9</th>
</tr>
</thead>
<tbody>
<tr>
<td>Flexion/extension</td>
<td>150°/−5°</td>
<td>70/0</td>
<td>100–120°</td>
<td>70–90°</td>
</tr>
<tr>
<td>Internal/external rotation</td>
<td>±6°/±30°</td>
<td>±10°</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Abduction/adduction</td>
<td>0–10°</td>
<td>0°</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Rollback (M/L)</td>
<td>5/15 mm</td>
<td>8</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Figure 5  Vertical tibio-femoral knee reaction force used for simulation of gait in wear simulators.
largest when the knee is fully extended. The tibial plateau is also sloped anteriorly and the combination of these two factors pull the collateral ligaments taught at extension. The result of the bony structures and taught ligaments creates a structure that is rotationally very stable at heel-strike.

An additional factor to consider at heel-strike is the translational stability of the knee. When the foot contacts the ground a natural anterior translation of the femur with respect to the tibia occurs. Anterior translation of the femur is restricted by the posterior cruciate ligament which prevents forward dislocation of the bearing. Unlike the hip, stability in the knee at heel-strike is, therefore, provided by a combination of bony structures, ligaments and muscle forces.

As walking progresses into mid-stance the knee begins to flex due to actions of the hamstrings muscles and the femoral condyles begin to roll on the tibial surface. This produces a natural posterior translation of the contact with the tibia along with an external rotation (5°) of the knee since the lateral femoral condyle has a larger radius and rolls further. The knee flexes to 20° (hamstrings) and then extends (quadriceps) back to 0°. At the change in direction from flexion to extension a second loading peak occurs in the knee as the muscle forces act to stabilise the joint. During mid-stance the knee is less stable since the medial/lateral spacing of the femoral condyles is larger and they no longer lock against the intercondylar notch. In addition to this, the radius of curvature of the femoral condyles during flexion is decreasing, and the natural rollback in the knee is pushing the contact down the posterior sloping tibial plateau which brings the femur closer to the tibia and increases the laxity in the ligaments. This laxity allows relative rotation between the femoral and tibial surfaces to occur.

As walking continues the knee flexes once again. Rolling occurs up to ~20° flexion until at which time the posterior translation of the femur relative to the tibia that occurs during the rolling action is restricted by the anterior cruciate ligament which acts like an anchor preventing posterior translation. As flexion continues and the condyles can no longer roll backward on the tibia the motion between the femoral and tibial surfaces changes from rolling to sliding. This change occurs initially with the medial condyle and then the lateral condyle resulting in a natural external rotation of the knee. At toe-off (~50° flexion) the quadriceps calf muscle (gastrocnemius) act to both stabilise the knee joint and produce plantar flexion of the ankle joint which accelerates the body forward resulting in a vertical joint reaction force of 2–4 times BW. A recent paper by Freeman and Pinskerova is recommended for further reading.

**Knee replacement**

Much like the hip correct positioning of the components during knee replacement is also vital in its success. The general alignment issues of the hip all hold true for the knee with even greater importance since the knee has a less conforming geometry and high stresses can lead to accelerated wear and delamination. Among the important factors for knee replacement positioning include rollback, tibial position/size, varus/valgus positioning and lift-off.

**Rollback**

The posterior shift in the contact between the femur and the tibia during flexion (rollback) increases the moment arm and efficiency of the quadriceps muscle. This is outlined in the schematic of Fig. 7. As flexion occurs, such as during a squat, the efficiency of the quadriceps muscle force will be directly related to tension and moment arm (PT) of the

![Figure 6: Motion of the knee used for simulation of gait in wear simulators.](image)
patella tendon. Clearly, without rollback the moment arm is short and a larger force is required by the muscle to produce the same action.

While rollback occurs naturally in the healthy knee, replacement joints are sometimes designed with a cam arrangement to encourage natural motion. Knee biomechanics following total knee replacement has been shown to be highly variable in clinical studies, with great patient variability.14-17

Tibial positioning/size
Incomplete support by the tibial cortex (cortical bone) for the tibial tray may lead to subsidence of the tibial tray if cancellous bone quality is poor.18 Generally, anterior medial and posterior lateral coverage is recommended when sizing components to prevent subsidence. An example of subsidence is shown in Fig. 8. The tibial insert in this case was lateralised as the patient had a valgus deformity and a para-

patellar arthrotenomy was conducted. This left very poor quality cancellous bone to support the medial side which subsided. Complete coverage is difficult due to the anatomical nature of the tibial plateau which extends further posteriorly on the medial side.

Overhang of the tibial insert is equally as important as undersizing. In some areas an overhanging tray can lead to impingement with the ligaments and tendons surrounding the knee and cause discomfort, pain and the need for revision.

Varus/valgus positioning and lift-off
In the natural knee, it is generally considered that 60% of the load is transferred through the medial condyle while 40% is transferred through the lateral condyle. In a knee with valgus deformity, the biomechanics and of the knee have changed significantly so that the majority of the load is transferred through the medial condyle. In contrast in a knee with a varus deformity, the majority of the load is transferred through the lateral condyle. Following knee replacement the loading on the tibial insert should be restored to as normal as possible. Edge loading of the polyethylene tibial insert caused by a varus/valgus deformity or by condylar lift-off has been shown in in vitro studies to cause accelerated wear of the polyethylene and should, therefore, be avoided.19,20

Joint loading in the hip and knee
There are two common methods for measuring the load acting on our joints. Inverse dynamics uses simple engineering mechanics as illustrated in Fig. 2 in the application of the body. The second method of measuring joint forces is to utilise instrumented prostheses.22 These are custom designed implants which contain complex instrumentation that

Figure 7  Schematic of the effect of rollback on the quadriceps moment arm (PT).

Figure 8  Medial tibial collapse (left) of a lateralised tibial component. Typical coverage of a tibial tray (dark line) and average patient data (right).
directly records the load acting on the joints in-vivo. Ethical approval is required for the use of these devices. Typical loads measured within the body during various activities for the hip and knee are shown in Table 3.

Kinesiology (inverse dynamics) has a tendency to overestimate joint forces due to assumptions related to the actions of muscles. In contrast, instrumented prostheses may produce more accurate results; however, there is a vast difference in the biomechanics from patient to patient. Therefore, to achieve an average value for a given population the sample must be very large to overcome the variability. The results from inverse dynamics and from instrumented prostheses are reasonably close and as long as you understand the limitations used in the analysis—either method is a very useful and valuable tool.

**The spine**

The spine, comprising three joints at any level—a disc and two facet joints, is arguably the most complex and demanding of any joint systems within the human body. Harms and Tabasso noted the importance of restoring the normal biomechanical environment as far as is possible during a surgical intervention and have proposed four key biomechanical functions for the spine (Table 4).

The key feature of this list is that the functions are listed in ascending order of importance and with the clinician give a set of principles by which an intervention should be approached. For example, the overriding concern is protection of the spinal cord which takes precedence over other considerations. If the functioning of the spinal cord can be assured then the stability of the spine should be the next important consideration as is the case in fusion surgery. This is achieved at the expense of the segment’s mobility.

**Loads in the spine**

As previously noted the spine, particularly the lumbar segment, experiences arguably the most demanding biomechanical environment of any joint system in the human body.

<table>
<thead>
<tr>
<th>Activity</th>
<th>Reference</th>
<th>Hip load</th>
<th>Knee load</th>
</tr>
</thead>
<tbody>
<tr>
<td>Walking</td>
<td>Freeman and Pinskerova</td>
<td>3.4 BW</td>
<td>3 BW</td>
</tr>
<tr>
<td></td>
<td>Paul</td>
<td>3–4 BW</td>
<td>4.4/4.9 BW</td>
</tr>
<tr>
<td></td>
<td>Bergmann et al.</td>
<td>2–3 BW</td>
<td>2–3.5 BW</td>
</tr>
<tr>
<td></td>
<td>Bergmann et al.</td>
<td>3–4 BW</td>
<td>3.5–5 BW</td>
</tr>
<tr>
<td>Stair ascent/descent</td>
<td>Bergmann et al.</td>
<td>2–3.5 BW</td>
<td>3–6 BW</td>
</tr>
<tr>
<td></td>
<td>Bergmann et al.</td>
<td>3.5–5 BW</td>
<td>4 BW</td>
</tr>
<tr>
<td>Rising from a chair</td>
<td>Ellis et al.</td>
<td>1.75–2.25 BW</td>
<td>3.2 BW</td>
</tr>
<tr>
<td></td>
<td>Bergmann et al.</td>
<td>4.7–5.6 BW</td>
<td>4.2 BW</td>
</tr>
</tbody>
</table>

This is because of the high loads that must be sustained, and the complex neuromuscular control that is required to maintain a stable yet mobile unit. The limits of the compressive loads within the spine are defined, principally, by the axial strength of the individual vertebrae (Fig. 9) that rise from approximately 1300 N at C3 (the third cervical vertebra) to over 8000 N at L4 (the fourth lumbar vertebra). Whilst a considerable margin of safety is built into these failure strengths the loads observed in the spine are often several times BW. The only time the compressive load is less than BW is in mainly the prone position.

The compressive forces arise largely from the muscle action that produces a counterbalancing moment to the weight of the upper torso and/or head that acts forward of the spine (Fig. 10) in a manner similar to the example given in Fig. 1. The posterior muscles have a relatively small moment or lever arm (b is typically 5 cm or less) and, therefore, have to produce a considerably larger force than the weight of the upper torso to produce a counterbalancing moment. The compressive load on the spine at that level is just the addition of the weight of the upper torso and the force generated by the posterior muscles. The axes of rotation of a given functional spinal unit, which is the disc and the adjacent two vertebrae, are generally located in or just below the disc, but the exact position will vary according to the type of motion being undertaken, the position of the spine and the nature of the individual as well as the functional spinal unit being observed. This simple figure can be amended to include a person carrying a weight in front of them. In this case, the posterior muscles will have to counterbalance an additional torque and therefore produce an even greater force to resist the forward flexing moment.
During level walking, the peak compressive forces developed across L4–L5 fluctuate between 1.5 and 3 times BW, and they do so at a frequency twice that of the gait cycle (Fig. 11). Impressively high transient loads (5–6 times BW or greater) occur for more extreme forms of activity or at the extremes of motion, with trunk flexion and/or high external loading being especially demanding. However, substantial loads are sustained even during 'inactively' standing or sitting (1.5 and 2.0 times BW, respectively). Also of importance are the observations of the shear forces in the lower lumbar spine. These loads are a source of much debate, in particular, on whether total disc replacements (TDRs) should be constrained or not, together with the effects these design features have on the adjacent structures, in particular, the facet joints. The important antero-posterior shear loads, which have been inferred from indirect measurement, can exceed 140 N during normal walking to more than 1000 N in more extreme activities. When considering any surgical intervention the clinician must also ensure that the forces and moments are distributed physiologically between the different structures within the spine with special reference to the facet joints. Typically, in the neutral position, 80% of the compressive load passes through the vertebral bodies and 20% through the facets in the lumbar spine, whilst in the cervical spine a greater proportion passes through the anterior column.

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Spinal motion is difficult to measure due to the segmented nature of the spine and the exceedingly high number of degrees of freedom together with elevated degree of redundancy that this allows. This must be coupled with issues of accessibility that may preclude routine non-clinical radiographic imaging of the spine with which to observe the behaviour of individual functional spinal units. Segmental motion assessment is more common in which surface measurement type devices are used. However, these have a number of shortcomings in that they are prone to the usual uncertainties found to be common with the use of surface markers but also the fact that the translational motion is often excluded from the analysis, which suggests that such motion is unimportant within the spine.

Observations show that the overall range of motion in flexion/extension varies enormously throughout the spine and varies within anatomical segments as well as between them. Combined flexion–extension is relatively high in both the cervical and lumbar regions exceeding 10° whilst a minimum is observed in the upper and mid-thoracic regions. The range of motion for lateral flexion is more even throughout the spine with values typically between 5° and 10°. The lower thoracic and lumbar spines are characterised by a relatively small range of motion in axial rotation which is typically of the order of 3°, which results from the
orientation of the facets which hinder this type of motion. During gait, the peak-to-peak flexion/extension motion in the lumbar spine increases with strenuousness of cadence, although the range utilised (typically 3–4°) remains only a modest fraction of that fully available. As noted previously, further complexity arises from the fact that the motions of the lower lumbar spine are kinematically coupled: most significantly, flexion/extension results in translational motion, thus causing the instantaneous axis of rotation to migrate. These coupled translations are of the order of 1–2 mm for L4–L5 and 0.5–1 mm for L5–S1, but they are highly variable due to the differences in functional spinal unit recruitment patterns occurring in forward versus backward trunk bending between individuals.

References