A variety of hip fracture fixation devices and techniques have been evaluated at the Hospital for Joint Diseases (New York, New York) using a number of biomechanical tests. As a result, factors have been identified that should be considered in interpreting results and clinical applicability of this type of testing. These test parameters represent tradeoffs between testing complexity and simulation of clinical reality. The effects of the following factors and their relative importance to results and conclusions will be reviewed: 1. test model; 2. fracture type and method of creation; 3. testing method type of support, test angles, load application, effects of cycling; and 4. methods of measurement and data analysis (experimental design and statistical analysis).

**Test Model**

Test models for biomechanical evaluation of hip fracture fixation include synthetic, fresh, or embalmed specimens. Fresh human specimens are the standard, as they best represent the in vivo condition of the bone. For embalmed and fresh specimens, anteroposterior and lateral radiographs of the proximal femur should be performed to rule out pathologic lesions or osseous defects. If possible, a history should be obtained to exclude patients with a history of metabolic bone disease or malignancy, since lesions in the proximal femur that are not radiographically apparent can affect specimen integrity. Since the bone density has been shown to be directly related to fixation strength and stability, it is necessary to use specimens with a similar bone density. Selection of specimens with a similar age range and sex can help to limit the variability in bone density. Dual energy x-ray absorptiometry (DEXA) scanning has been shown to effectively and reliably quantitate the degree of osteoporosis in specimens.

This is done by the measurement of bone mineral density (BMD) at either the lumbar spine or the proximal femur (intertrochanteric and femoral neck regions), as standardized BMD values have been developed for these sites. Computerized tomography has also been shown to be an effective tool in determining proximal femoral bone density but can be costly to perform. The use of Singh indices has been shown to be a poor indicator of proximal femoral bone density and these indices have a low interobserver reliability.

For fresh specimens, testing should be completed in several days, as the specimens may decompose, affecting their biomechanical characteristics. Care should be taken to avoid desiccation, as this will also affect its biomechanical characteristics. To minimize the effects of specimen variability, the instrumented specimen’s stiffness values can be normalized to the stiffness of the intact (control) specimens. The use of fresh specimens also poses difficulty with transducer (i.e., strain gauge) mounting. Specimen acclimation to room temperature after refrigeration or freezing is necessary, as temperature may affect the specimen’s mechanical characteristics and will affect strain gauge measurements. Refrigeration of specimens between experiments will improve specimen longevity. Also, the possibility of disease transmission (HIV, hepatitis) should be considered.
Lastly, limited availability and cost make the use of fresh specimens a luxury that some laboratories cannot afford.

With the use of embalmed specimens, as with fresh specimens, particular care should be taken to avoid desiccation. Blanton and colleagues have shown that there is no difference in the bone density of embalmed specimens compared to fresh bone. Evans and associates and Wainer and colleagues also demonstrated no differences between the mechanical properties of embalmed and unembalmed long bone specimens. However, McElhaney and associates demonstrated that embalming causes a significant reduction (12%) in the ultimate compressive strength of bovine bone. Slight reductions were seen in the ultimate tensile strength, modulus of elasticity, and maximum strain. As a result, when embalmed specimens are used for biomechanical testing, they may behave mechanically in a more osteoporotic manner than their densitometry measurements indicate. Lastly, it is unclear how embalming effects HIV and hepatitis transmission. Nonetheless, universal precautions should be used when handling embalmed specimens.

The effect of embalming on soft tissue must also be taken into consideration when soft tissue is incorporated into the fracture fixation. Depending on the quality of embalming, the soft tissues may develop a “leathery” quality. This becomes a factor when soft tissue becomes incorporated into the repair, such as with tension band wiring of the greater trochanter fractures where the abductor tendon is included with the repair. This may alter the experimental results since fixation strength is ultimately dependent on the quality of the soft tissue, which is often of poor quality in embalmed specimens.

To minimize the effects of variation in embalming techniques, degree of desiccation, and other uncontrollable factors, construct stiffness can be normalized to the stiffness of the intact (control) specimens. Strain data and displacements should also be similarly normalized. In addition, Reikeras and colleagues demonstrated in matched, paired cadaver femurs that the anatomy of one femur is not equivalent to that of the contralateral side in the same individual. Swiontkowski and associates demonstrated that left proximal femur specimens were approximately 5% stiffer than right femurs of matched pairs. This may be attributable to the significant difference in femoral anteversion between matched pairs. However, Mather demonstrated no left-right differences in ultimate load, modulus of elasticity, and extreme fiber-breaking stress, but energy absorbing capacity was slightly higher on the left-sided specimens. Despite these conflicting results, randomization to ensure equal number of left/right specimens between groups is necessary.

Synthetic specimens consisting of epoxy composite (Sawbones, Pacific Research Laboratories, Inc., Vashon, WA), in contrast to embalmed and fresh specimens, have standardized density measurements with reproducible known mechanical properties. Swiontkowski and colleagues suggested the use of synthetic bones when performing biomechanical analysis of hip fracture fixation to limit the variability in both proximal femoral anatomy and bone density between matched pairs of embalmed or fresh specimens. Other considerations for synthetic specimens include a difference in screw purchase between synthetic specimens and embalmed/fresh specimens, since the composite structure differs from cortical and cancellous bone. Also, the simulated fracture surfaces lack the interdigitation seen with the simulated fractures created in embalmed or fresh specimens.

**Fracture Type and Method of Creation**

The standard technique to recreate in vivo fracture patterns includes cutting only the cortical bone with a sagittal saw and completing the fracture with a mallet blow(s). This technique creates an irregular fracture surface with interdigitation of the fracture fragments on fracture reduction. A variation of this technique involves drilling multiple holes along the intended fracture line and completing the fracture with a mallet blow. Both of these techniques attempt to recreate the interdigitation of clinical fractures with fracture reduction. However, one must ensure that the fracture pattern is easily duplicated and that they have a similar degree of interdigitation. Placing the specimen in a material testing system or dropping a weight on an intact specimen can create simulated fractures in intact femurs. This is problematic, however, because of the difficulty of controlling the location and number of fracture fragments as well as consistently producing a reproducible fracture pattern. Swiontkowski and colleagues reported that the creation of a simulated fracture by means of an osteotomy with a saw is the worst-case scenario for in vitro mechanical testing because of lack of interdigitation of the fracture surfaces that would normally occur in vivo and provide fracture stability. However, the benefits include easy reproducibility and standardization of the fracture pattern.

**Femoral Neck**

Klennerman and associates demonstrated that femoral neck fractures have a consistent fracture pattern. They examined 20 femoral heads removed after fracture and showed that the femoral neck fracture occurs most commonly in the subcapital region. Clark and colleagues and others have reproduced this fracture pattern when biomechanically evaluating femoral neck fractures. The subcapital region, immediately below the femoral head is identified and the cortex is scored circumferentially with an oscillating sagittal saw. The fracture is completed with a mallet blow(s) to the femoral head. Some investigators, rather than performing a subcapital osteotomy, have performed other types of femoral neck osteotomies to evaluate femoral neck fracture fixation. Some have used a transcervical osteotomy.
vertically oriented osteotomies,\textsuperscript{20,21} and basocervical fracture pattern.\textsuperscript{59} Because these femoral neck fracture patterns are rare, their replication for biomechanical testing may lead to erroneous results.\textsuperscript{14}

To evaluate femoral neck fractures with posterior comminution, Goodman and colleagues\textsuperscript{22} performed a standardized osteotomy through the femoral neck with an oscillating saw, bisecting the distance between the lower cartilaginous portion of the femoral head and the intertrochanteric line. A 5-millimeter thick slice of bone was then excised from the posteromedial quadrant of the distal fragment. Recent biomechanical investigations at the Hospital for Joint Diseases by Kauffman and associates\textsuperscript{23} analyzed the effects of posteromedial comminution on fracture fixation. Simulated subcapital femoral neck fractures were created by cutting the circumference of the cortical bone with a thin-blade reciprocating saw just distal to the end of the articular cartilage. The fractures were completed with a mallet blow to simulate the jagged surfaces seen clinically. A simulated posterior comminution defect (1 cm wide by 1.5 cm long by 0.5 cm deep) was then created. Both are valid techniques to reproduce and test femoral neck fractures with posterior comminution.

**Intertrochanteric Fractures**

Stable intertrochanteric fracture patterns have been investigated by numerous authors.\textsuperscript{24-27} The standard technique is to partially cut the cortex at the anterior capsular insertion (intertrochanteric line) with an oscillating saw. The cut extends distally through the medial femoral cortex across the tip of the greater trochanter and continues posteriorly along the femoral capsular insertion. The cut is completed with a blunt blow to simulate the jagged surfaces seen clinically.

For unstable intertrochanteric fractures, Kaufer and colleagues\textsuperscript{28} described a technique for the creation and testing of unstable intertrochanteric fractures. The fracture pattern is initially identical to the stable intertrochanteric fracture pattern (Fig. 1). Two additional saw cuts create “free” greater trochanter and posteromedial fragments. The creation of the posteromedial fragment is an attempt to recreate the clinically seen situation of loss of the posteromedial buttress. This is the key element in unstable intertrochanteric fracture patterns. Other investigators have also produced similar fracture patterns in their evaluation of unstable intertrochanteric fractures, varying the size of the posteromedial fragment.\textsuperscript{24-26,29} We believe this is crucial for the testing of simulated intertrochanteric fractures as this attempts to replicate in vivo events.

**Subtrochanteric fracture**

Tencer and colleagues\textsuperscript{30} described a technique for creation of stable and unstable subtrochanteric fractures. The stable subtrochanteric fracture model requires a complete transverse saw cut that is three centimeters distal to the greatest prominence of the lesser trochanter. The subtrochanteric region is predominantly cortical bone and thus, fracture interdigitation is minimal. As a result, an osteotomy is preferred over a mallet blow, as this will limit variability in the fracture pattern. The unstable segmental fracture model was created by making a second transverse cut three centimeters distal to the first cut and six centimeters distal to the lesser trochanter. The segment of proximal femur was then discarded, creating an unstable segmental subtrochanteric fracture. This fracture model, allowing full bone-to-bone contact, in the stable configuration and negligible bone-to-bone contact of the unstable setting encompasses the range of in vivo mechanical conditions required to evaluate fixation systems for subtrochanteric fractures. Other biomechanical studies have also employed this fracture model when evaluating fixation devices for subtrochanteric fractures.\textsuperscript{31-34}
Other fracture patterns have been described that are variations of the Tencer and associates\(^{30}\) model.\(^ {35-39}\) Wheeler and colleagues\(^ {35}\) used a 30-degree wedge osteotomy, removing the lesser trochanter to simulate a high, subtrochanteric femur fracture with loss of the medial column but with preservation of the piriformis fossa (Fig. 2). While many simulated subtrochanteric fractures have been described, it is clear that the investigators attempted to duplicate the wide array of clinically occurring fracture patterns. The Tencer and associates\(^ {30}\) fracture pattern is the easiest to reproduce. Oblique fractures, which have variations in angulation that affect inherent fracture stability, are difficult to replicate experimentally. It is difficult and time consuming to create wedge osteotomies without the use of a template, which itself has inherent problems of reproducibility depending on template positioning.

**Testing Method**

**Test angles**

The orientation of the joint reactive force at the hip varies during the gait cycle, ranging from 12.5 degrees to 25 degrees from the vertical.\(^ {40}\) This is calculated by adding the angle of the normal inclination of the femoral shaft during one-leg stance, which is 7 to 10 degrees,\(^ {41}\) and the angle between the resultant force on the femoral head, 10 to 15 degrees from the midline.\(^ {42,43}\) Stresscoat brittle coatings and strain gauge analysis have demonstrated that a femoral shaft inclination of 20 to 30 degrees, simulating forces during one-leg stance, causes a more even distribution of strain in the femoral neck than with more vertically applied loads.\(^ {44}\) Stresscoating involves the application of lacquer to the proximal femur, which upon application of loads will fracture at sites of increased stress concentration. Based on this, Frankel\(^ {44}\) determined that the optimal loading angle for in vitro testing of hip fractures is 17 to 25 degrees from the vertical. Furthermore, he demonstrated that varying the direction of the force changed the strain distribution in the proximal femur.

Oh and Harris,\(^ {45}\) on the other hand, demonstrated that the differences in proximal strain, as the angle of load application changed from 0 to 20 degrees, was not significant. However, there was a trend for decreasing compression medially and decreasing tension laterally as the angle of adduction increased from 0 to 20 degrees. Chang and colleagues\(^ {34}\) demonstrated that femurs loaded at 0, 10, and 20 degrees of femoral shaft adduction demonstrated no significant differences in lateral and medial strain distribution. Cochran and associates,\(^ {37}\) however, demonstrated that in 0 degrees of femoral shaft adduction, strains increased by a mean factor of 1.8 for tension and 1.2 for compression, compared to femoral shafts tested at 10 degrees of adduction. They, as did Frankel,\(^ {44,46}\) stressed the importance of correct shaft position during mechanical testing to replicate in vivo forces, since changes in femoral shaft position by as little as 10 degrees will alter strain distribution in the subtrochanteric region. This was confirmed by Frankel\(^ {44,46}\) in studies of the femoral neck who demonstrated that changes in the amount of adduction of the femoral shaft considerably affected strain distribution in the femoral neck.

Despite this conflicting data, the authors have attempted to replicate and recreate the resultant forces on the femoral head seen in vivo. There appears to be a change in the tension/compression forces seen medially and laterally, which may play a significant role in fixation failure. The femoral shaft should be placed in 17 to 25 degrees of adduction and neutral version, as was originally described by Frankel,\(^ {44,46}\) to accurately recreate the joint reaction force. Correct shaft inclination is confirmed by the use of a goniometer.

Prior to loading, the femoral condyles are removed and the distal 5 centimeters of the specimen are potted in fixtures with either methylmethacrylate bone cement, Wood’s metal, epoxy auto body filler, or any other substance that will provide a secure fit and have sufficient strength. The degree of shaft inclination is adjusted by the use of a vise (Fig. 3). Others\(^ {34,35}\) have mounted the rectangular potting fixture to a universal joint, which allowed the femur to automatically attain an alignment of 17 to 25 degrees from the vertical and simulate the combined axial and bending loads of single-leg stance. This is acceptable as long as a reproducible femoral shaft inclination is produced. The first specimen is not included in the experiment and functions to calibrate testing machines and assure test reproducibility.
Hip joint forces have been calculated to range from a minimum of body weight during midstance to a maximum of eight times the body weight near toe-off in fast walking. Pauwels47 calculated that during two-leg stance, ignoring muscular and ligamentous factors, the force at the femoral head is one-half of the body weight. During one-legged stance, this force increases to three times the body weight. Others have reported similar results.42,48 Osborne and Fahrni 49 attempted to determine forces at the hip by including the abductor musculature. Their model consisted of the pelvis and femur with the hip joint containing a fluid-filled rubber bag connected to a manometer. They measured the pressure passing through the joint under various weights and gluteal muscle pull. They determined the joint pressure was the summation of the body weight and gluteal pull.

Frankel50 demonstrated in a cadaver model that forces simulating muscular and ligamentous attachments increase the load on the femoral head during single-leg stance. Springs and wires were attached to the pelvis and femur to simulate the checkrein effects of the short rotator muscles and the iliotibial band. Cochran and colleagues77 developed an in vitro model to determine the effects of the abductors and iliotibial band in the subtrochanteric region during femoral head loading. Testing was done with the femoral shaft aligned at 10 degrees of adduction and neutral in the anterior-posterior plane. The action of the abductors and tensor fascia lata was simulated with cables attached to the greater trochanter (Fig. 4). They found that with the addition of the abductors, tension strain increased by a factor of 6.6, and compression strain increased by a factor of 4.5. The addition of the tensor fascia lata with the abductors resulted in an increase in the tension and compression strain by a factor of 3.3. As the tension in the tensor fascia lata cable was increased, strain decreased in the proximal femur, apparently neutralizing the effects of the abductors. The addition of abductor and tensor fascia lata forces adds significant complexity to the experiment and may interfere with the fixation technique that is under investigation, especially for femoral neck and intertrochanteric fractures. It also adds an additional variable, which has to be controlled during the experiment. However, the simulation of these forces produces a situation that more closely approaches the forces experienced by the proximal femur in vivo.

The authors believe that the muscular forces take on considerable importance for subtrochanteric fracture fixation since the abductor musculature is attached to the greater trochanter and has a clinically significant effect on fracture displacement, as demonstrated by Cochran and colleagues.37 We once used a similar testing apparatus that employed 18-gauge trochanteric wires to simulate abductor forces and displacement gauges connected to a shaft collar located below the fracture. However, since this method utilized a lever-arm system, it could not achieve abductor forces.
greater than femoral head loads. Lastly, it may be difficult to consistently replicate the effects of the tensor fascia lata. Therefore, rather than include it in testing as an unreliable variable, the authors decided to exclude it to minimize variability in the data, realizing this is not ideal.

**Load Application**

The chief criticism of static loading and loading to failure is that it does not simulate the in vivo cyclic weight-bearing pattern of the proximal femur and, thus, has little clinical application. Cyclic testing attempts to estimate the effect of an extended period of weightbearing that simulated months to years; however, the information can be misleading as it does not truly represent what occurs in vivo. It does not take into account the effects of bone remodeling and healing, which would increase fracture stability. Bone remodeling, which occurs during in vivo loading, cannot be produced in test specimens. However, despite this, cyclic loading can investigate the effects of short periods of weightbearing on implant stability. At the Hospital for Joint Diseases, specimens are initially statically tested then cycled for 10,000 cycles at 1 Hz to determine the effects of an extended period of weightbearing. Other investigators have also chosen a similar loading regimen of 0 to 3 times the body weight acting sinusoidally at 1 Hz. This simulates the forces experienced during single-leg stance and is within the peak theoretical load acting on the hip joint.

Loading rates to bone have been shown in a number of studies to be an important factor affecting the ultimate strength of bone. Raftopoulous and associates demonstrated that increases in strain rate resulted in increased stiffness of proximal femur test specimens. Stiffness of the specimen approximately doubles as the rate increases (0.001 to 100 seconds). However, properties of fixation devices (e.g., metal) do not exhibit this effect. The goal of testing is to perform testing within a reasonable time, at physiological loading rates (0.1 to 1.0 seconds).

Load application to the femoral head should duplicate clinical reality. Kaufer and colleagues attempted to recreate loads through the acetabulum by utilizing a large metal cup lined with hard rubber over the femoral head to produce an even load distribution. Goodman and associates transferred loads to the femoral head through cement that profiled the femoral head. Others delivered loads through a shallow molded polypropylene cup or plastic cup with sufficient contact area to prevent local damage to the femoral head. Some, rather than attempting to create a uniform surface for load application to the femoral head, applied loads through the native acetabulum. Loads applied through a cup or cement profiled to the femoral head will uniformly distribute forces. The ideal situation would be to apply loads to the native acetabulum. Cost, difficulty with obtaining specimens, and difficulty with load application through the native acetabulum (securely holding the pelvis) make this option unrealistic.

A variety of other devices have been described that impart loads to the proximal femur for laboratory tests, which may not be as reliable as loads applied through devices profiled to the femoral head. Ramser and colleagues used a custom-made jig fabricated from hardened steel in which a four-pin grip held the proximal femur against a piece of half-pipe. The half-pipe abutted the lesser trochanter to prevent migration of the femur within the jig. At the Hospital for Joint Diseases, a polished flat applicator is utilized to apply loads to the femoral head which permits free movement of the femoral head. Other investigators have also utilized this technique. Limitations include the nonuniform distribution of force to the femoral head, as seen in the native acetabulum, and the possibility of the femoral head sliding beneath the flat plate, altering the load/deformation curve and underestimating stiffness. If independent displacement gauges are not used, displacement of the femoral head will be affected by bending of the femoral shaft, erroneously reading bending of the femoral shaft as motion at the fracture site.

The usual types of tests applied in biomechanical testing of hip are axial, bending, and torsion. These three loading schemes have been used in numerous biomechanical studies as they are the primary in vivo loading conditions. Axial loading represents the physiologic loads seen during weightbearing, and bending loads are seen as a patient raises himself from a chair. Torque represents rotational loads caused by friction between the femoral head and acetabulum, which are generally low in the hip. At the Hospital for Joint Diseases, during axial loading studies, the femoral shaft is positioned in 20 to 25 degrees of adduction in the coronal plane and is neutral in the sagittal plane (see Fig. 3). Loads are applied continuously over 1 to 5 minutes or in 100 Newton increments to 1,200 Newtons to simulate one-legged stance (approximately 2.5 times body weight). Lateral bending is performed with the femoral shaft and neck held parallel to the testing platform by a vise; the femur is supported at the level of the lesser trochanter by a steel gauge block to minimize posterior bending of the femoral shaft, thus preventing misleading displacement values. Vertical loads are applied to the anterior aspect of the femoral head perpendicular to the femoral neck. During torsional testing, the diaphysis is fixed at an angle to achieve a vertical neck orientation. The femoral heads are fitted with a steel hose clamp to avoid slippage and then clamped in a three-jaw chuck. A 100 N axial preload is applied to the femoral head after osteotomy to standardize inter-fragmental contact, and the femoral head is rotated clockwise at 15 degrees/minute until failure or to generate a load/deformation
curve to determine torsional rigidity of the construct. Swiontkowski and associates described a torsional testing apparatus that consisted of the femoral head gripped at the equator with 12 radial pins inserted to a depth of about 2 millimeters. The femoral shaft was held 3 centimeters below the lesser trochanter in a six-pin grip (Fig. 5). The authors used a Tinius Olsen torsion testing machine which applied an 890 N compression preload to standardize inter-fragmental contact at the osteotomy site. The shaft was rotated around the neck axis at a rate of 270 degrees/minute. Baril and colleagues utilized a similar testing apparatus for torsional testing of subcapital femoral neck fractures but rotated the femoral head at a rate of 50 degrees/minute. At the Hospital for Joint Diseases, torsional testing is rarely used since it is felt that a pure torsional moment is not experienced by the hip in vivo. Thus, the clinical applicability of such data is questionable.

Measurement Method

Displacement

Displacement gauges are employed to measure displacement of the femoral head in response to loads. At the Hospital for Joint Diseases, four displacement gauges are placed orthogonal to the head and shaft. Movement at the fracture site in the anteroposterior and superoinferior planes are recorded. Displacements gauges connected to the femoral shaft take into account bending of the femoral shaft, which may be incorrectly interpreted as motion at the fracture site. Currently at the Hospital for Joint Diseases, one displacement gauge is connected to a ring surrounding the proximal femoral shaft measures inferior head displacement since previous investigations displayed minimal motion in the anteroposterior plane. This also takes into account the bending displacement of the shaft some investigators have mistaken as movement at the osteotomy site.

Other techniques have been described to measure displacement during in vitro loading. Van Audekerke and associates used an image intensifier connected to a videotape system that recorded the exact position of the femoral head and fixation device during loading. It allowed playback of the events during loading with the possibility of freezing an image at any time. Problems include assuring correct rotation of the specimen, as this will effect the perceived degree of displacement. Elmerson and colleagues measured motion at the osteotomy site with two highly compliant displacement gauges which were constructed by welding needles onto spring steel hemicircles (Fig. 6). The strain gauge elements were connected to a constant current bridge, and the variations in the electrical signal, caused
by displacement of the femoral head during internal fixation, were amplified, filtered, and recorded on an oscilloscope. Engesaeter and associates assessed displacement at their osteotomy site by mounting an electronic extensometer on two parallel metal pins drilled into the bone on each side of the osteotomy. Both these techniques require extended set-up time and may be affected by malfunctioning electrical gauges and oscilloscopes.

**Strain Measurement**

The measurement of strain (change in length of an object divided by its original length) in the proximal femur usually involves the application of strain gauges to the outer surface of the femur. Other techniques include laser interferometry, photoelastic models, and stress-coat techniques. These techniques are more costly and more labor intensive and are rarely utilized at the Hospital for Joint Diseases. Strain gauges should be placed longitudinal to the long axis of the femur, as based on the work of several investigators who determined, with rosette strain gauges, that the resolved principal strain direction was generally concomitant with the long axis of the bone in the proximal femur. Based on this work, unidirectional strain gauges can be applied to the proximal femur by the method described by Oh and Harris.

It has been determined that tests extending over long periods of time or in conditions where drying occurs quickly are subject to specimen shrinkage which significantly affects strain gauge data. Temperature compensation is unnecessary for short-term tests as long as ambient temperature variation is small. Refrigerated specimens must be acclimated to room temperature because gauges are temperature sensitive. It is also important to keep specimens moist throughout the experiment as moisture content of bone has been shown to significantly affect Young’s modulus values.

**Data Analysis**

The biomechanical evaluation of hip fracture fixation investigates the effects of various physiological forces on a fracture fixation construct. The applied force causes deformation and strain of the test specimen and displacement at the fracture site. Data can be retrieved in graphical form for load/displacement or by computer acquisition programs that generate load/displacement curves and perform calculations on the data. The stiffness of the construct, or fixation stability, is the slope of the elastic region of the stress/strain curve. Data is usually presented in Newtons/cm for axial and bending load or Newton/meters for torque. As the force applied is increased, the elastic limit of the construct is exceeded. At this “yield” point, some non-reversible failure of the construct occurs. It can be a screw cutting-out of the femoral head or deformation of the fixation device. Loading to failure and fixation stability, or stiffness, are two ways that a device can be characterized.

The experimental design should consist of a sufficient number of specimens to obtain statistically meaningful differences or similarities. This is accomplished by performing a power analysis. Most biomechanical experiments involve comparing one fixation method to another. Statistical analysis using the Student t-tests is sufficient in these cases. In the case where more than two fixation devices are biomechanically tested, a Student t-test is no longer valid. An analysis of variance (ANOVA), which determines whether a significant difference exists between two or more sample means, is usually performed.

**Conclusion**

In this review, an overview of the biomechanics behind various techniques used to analyze hip fracture fixation systems is presented. The ultimate goal of laboratory studies is to replicate clinical reality and apply knowledge from these studies to clinical practice. As has been demonstrated, it is important to carefully review the methods of these biomechanical investigations prior to widespread clinical application, as methodology can potentially bias investigations ultimately affecting patient care. Nonetheless, these biomechanical tests can give insight into potential problems that may arise with clinical use.

**References**

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