RELATIONSHIP BETWEEN ANKLE JOINT ANGLE AND GROUND CONTACT TIME IN ATHLETE DROP JUMP PERFORMANCE

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By

Joshua Phillips

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The thesis of Joshua Phillips is approved by:

William Whiting, Ph. D. Date

Shane Stecyk, Ph. D. Date

Konstantinos Vrongistinos, Ph. D. Date

Sean Flanagan, Ph. D., Chair Date

California State University, Northridge
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ABSTRACT

RELATIONSHIP BETWEEN ANKLE JOINT ANGLE AND GROUND CONTACT TIME IN ATHLETE DROP JUMP PERFORMANCE

By

Joshua Phillips

Master of Science in Kinesiology

The purpose of this investigation was to observe the influence of ankle joint angle and ground contact time (GCT) on jump height performance (JH). Eighteen male junior college football players performed depth jumps in four distinct jump conditions: dorsiflexed-fast (DFF), dorsiflexed-slow (DFS), plantarflexed-fast (PFF) and plantarflexed-slow (PFS). The four jump conditions were evaluated against the dependent variables GCT, JH, reactive strength index (RSI), ankle angle at contact (AAC), peak ankle flexion angle (PAFA), ankle range (AR), knee angle at contact (KAC), peak knee flexion angle (PKFA), knee range (KR), hip angle at contact (HAC), peak hip flexion angle (PHFA), hip range (HR), initial GRF and peak GRF. The fast conditions showed significant main effects for decreased GCT, decreased joint range for the knee and hip,
decreased peak flexion at all 3 joints, as well as increased initial and peak GRF (all p < .05). Contrary to our hypotheses, the foot condition was not related to a significant main effect for GCT or JH. However, the foot condition did result in a significant main effect for ankle angle at contact, ankle range, initial GRF and peak GRF. Neither the foot nor the speed conditions displayed a significant main effect for jump height. The results indicate that depth jump GCT can be reduced without a significant corresponding decrease in jump height. A ground contact time of 0.3 s may be the minimal GCT possible to elicit maximum jump height. However, depth jumps over 45 cm may elicit eccentric loads that are too great for subjects to achieve fast stretch-shorten cycles, less than 250 ms. Increased dorsiflexion of the ankle at ground contact does not significantly affect ground contact time or jump height, thus, it is not recommended at this time. Future research should revisit these four jump conditions at depth jump heights above and below 45 cm. Additionally, the longitudinal effects of depth jump training should be examined.
INTRODUCTION

Force, work and power are fundamental concepts in human movement and have been studied for centuries. In recent decades, the utilization of elastic energy has emerged as a critical component of movement. Research has revealed that the muscle tendon complex (MTC) is capable of storing elastic energy in its series elastic component (SEC), which can result in enhanced capabilities for mechanical work and power (52, 90). Elastic energy is stored within the SEC when a stretch is applied (50). This energy can be utilized for mechanical work and power through the stretch-shorten cycle (SSC). The SSC is characterized by an eccentric action immediately preceding a concentric action (102). The length of time taken to complete a SSC has surfaced as a major determinant of the contribution of elastic energy to mechanical work and power.

Research has focused predominantly on dynamic movements, jumping and running, to evaluate the involvement of elastic mechanisms in force production. McBride et al. (71) evaluated 3 jumping movements with increasing levels of muscle pre-activity and eccentric load before concentric action, comparing the static jump (SJ), counter-movement jump (CMJ) and drop jump (DJ), respectively. The data showed a significant increase in force output throughout the three jumps, escalating with MTC pre-activation (71). Eccentric muscle activity correlated highly to muscular stiffness. Arampatzis et al. (6) showed that individuals have different optimum stiffness levels based on the demands of the activity and that increased stiffness leads to shorter ground contact times (GCT). Decreased GCT will lead to amplified returns of elastic energy stored during the contact phase (6, 64, 102).
Stiffness of the MTC may enhance force transmission from muscular contraction through the tendinous system (102). The MTC will not transmit any force to the skeletal system until the slack is taken up from SEC (52). The time lag that exists between the increase in electrical activity in the MTC and the mechanical response of the muscle, segment movement, is called electromechanical delay (EMD) (52). Muraoka et al. (75) revealed that the EMD of the MTC increases if there is slack in the major tendon associated with concentric joint movement, thus increasing GCT and decreasing the use of elastic energy. For example, by increasing dorsiflexion at the ankle joint, slack will be removed from the Achilles tendon leading to decreased EMD in the gastrocnemius and soleus.

Despite the plethora of research that has explored the role of elastic energy in human movement over the last forty years, a void remains in the application of the findings. An in-depth analysis of the mechanisms behind elastic energy use and direct application of these mechanisms to dynamic athletic performance is still needed. This study examines elasticity through DJ performance using four jumping techniques designed to contrast pre-active and relaxed muscle, through ankle joint angle. The SSC was separated into fast vs. slow conditions evinced through GCT. The following outcomes are predicted: 1) the dorsiflexed conditions will produce shorter GCT and higher ground reaction forces (GRF) than the plantar flexed conditions 2) shorter ground contact times will evince a negative correlation to jump, and 3) the loss in jump height, attributable to dorsiflexion will not be significant enough to outweigh the athletic benefits of decreased GCT.
The literature review covers: 1) elasticity and the spring-mass model, 2) stretch-shorten cycle, 3) fascicle vs. series elastic component contributions to movement, 4) stiffness, 5) contribution of stiffness to human performance, and 6) stiffness control mechanisms, injury prevention and training applications.

**Elasticity and the Spring-Mass Model**

Elasticity has been defined as a measure of how readily a material will reform after it has been deformed by an external load (33). The elasticity of passive bodily tissues (tendons) combines with the active functioning of muscle fibers to control the progression of contractions. Hill (52) explained the function of musculotendinous structures in elastic energy usage via a 3-component model. The three components can be separated into a contractile component (CC) that exerts active force during muscle shortening and two distinct elastic components, the series elastic component (SEC) and parallel elastic component (PEC).

The CC is the result of cross bridge formation between thick myosin and thin actin filaments (33). Muscle contraction is the result of cross bridge formation within the myofibrils of the muscle fiber sarcoplasm. Cross bridges are formed when a myosin head bonds with an available actin molecule. This bond pulls the actin molecule towards the center of the sarcomere, in an event known as the power stroke. With the help of ATP, this connection is broken and the myosin head is free to bond with another actin molecule further along the actin filament. In this way, muscle fibers contract through the mechanism of thin actin filaments sliding along the thick myosin filaments toward the center of the sarcomere (20). The process of cross bridge formation can be explained in
more rigorous detail; however, for the purposes of this study the salient points relate to the force and velocity of the contraction. Force depends on the number of cross bridges formed between actin and myosin filaments. An inverse relationship exists between the maximum force generated and the speed of the contraction. The explanation for this inverse force velocity relationship again centers on cross bridge formation. To increase the velocity of sarcomere shortening the cross bridges must cycle more rapidly, which decreases the number of cross bridges formed leading to diminished force production (33).

The elastic elements of the musculotendinous complex, connected to the CC, influence muscle activity in two ways. The SEC and PEC segments are capable of storing elastic energy during stretch and the SEC segments may also affect changes in muscle-fiber length during both stretching and shortening contractions (20). The SEC has conventionally been thought to lie within the tendinous tissues. However, new evidence has put forth a model that places portions of the SEC within the muscle fiber as well. Elastic energy is stored within the SEC of the muscle fiber, when cross-bridges are formed and a stretch is applied (50). The PEC is comprised of the sarcolemma and muscle fascia and is responsible for force exerted by relaxed (passive) muscle when it is stretched beyond its resting length (52). However, the amount of stored mechanical energy by the PEC is minute, thus, its contribution to propulsive force is considered small (91). Hill’s model, commonly known as the spring-mass model, predicts that the slack in the SEC must be removed before force can be transmitted externally to attached segments (33).
The spring-mass model illustrates the spring-like properties of body segments and individual joints as springs. Compression of the spring, during ground contact, stretches the MTC, including the CC, SEC and PEC. Similarly, when the spring rebounds, during toe-off, the MTC shortens or contracts (23). The model also assumes that the center of mass (COM) achieve equivalent heights and velocities at landing and takeoff (15, 73). Blickhan (15) counters that this is an oversimplification because the COM will never achieve identical heights and velocities because of variations in leg stiffness. Special consideration of the SEC is appropriate because it is primarily responsible for storing energy between the cross bridges of sarcomere filaments (50). This energy is either used for subsequent concentric contraction or it dissipates as heat. The mechanism through which the recoil of the MTC spring after stretch produces increased capacity for work and power is referred to as the stretch-shorten cycle.

**Stretch-Shorten Cycle**

When tissue is stretched causing eccentric tension immediately prior to concentric contraction the SSC is initiated (102). In 1965, G. A. Cavagna (26) analyzed the work capacity of a frog gastrocnemius and found three times more production in the prestretched condition. Follow up research, by Cavagna et al. (24), was able to elicit comparable results in eccentrically stretched human forearm flexors. More recent investigations have examined mechanical measures of power and work in relation to sprinting performance (23, 25). Cavagna et al. (25) discovered that intrinsic properties of muscle contraction were responsible for increases in power output up to 5 meters per second. However, additional increases in power output required to reach maximal
velocity were ascribed to the storage and subsequent release of elastic energy in applicable leg MTCs. As recently as 2006, Cavagna (23) demonstrated the contribution of the elastic components of the MTC through the duration of mechanical work. The duration of positive external work was greater than the negative external work duration up to a velocity of 14 km· h⁻¹, 3.89 m· s⁻¹, suggesting a contribution of muscle fibers the change in length of the MTC. However, above that speed the durations of positive and negative external work were approximately equivalent, indicating that elastic components are almost entirely responsible for the length change.

Scientists are not in complete agreement on the role of elastic energy use in enhanced work output following a countermovement. Van Ingen Schenau et al. (93) asserted that the observable increase in work production can be rationalized through time to build up force development, without regard for the storage of elastic energy. They suggested that the eccentric phase of the SSC maximizes torque by providing more time to stimulate relevant muscle tissue, thus increasing cross bridge formation. Bobbert et al. (16) supported this hypothesis through research that demonstrated a direct correlation between vertical ground reaction force and the time to stimulate the gluteus maximus. Van Ingen Schenau et al. (93) also proposed that muscle fiber type predisposes individuals to the benefits of the SSC. Individuals with slow-twitch, Type 1, profit to a greater extent because they have slower cycling rates than fast-twitch fibers. Slow-twitch fibers need the increased time to build up force development provided by a countermovement, whereas fast-twitch fibers do not.

Contrary to these reports, several researchers have defended the role of elastic energy use in enhanced work and power production though the SSC (50, 106). Herzog
(50) accounted for the benefits evinced in individuals with predominantly slow-twitch fibers through the storage of elastic energy within filament cross bridges. The timing of the SSC is critical because it must occur before the cross bridges detach for the stored energy to be used for work. Slower cycling rates of the slow-twitch fibers increase the likelihood of contribution from stored elastic energy in the SEC. Zatsiorsky (106) also questioned the findings of Van Ingen Schenau et al. (93) for failing to elucidate the mechanisms through which athletes execute take offs of much shorter duration 300-500 ms, including sprints, high jump and long jump. Fini et al. (39) added more compelling evidence against the time to enhance force production as a comprehensive explanation of the SSC. They demonstrated that vertical jump performance was significantly improved when proceeded by a countermovement as opposed to a buildup of maximal isometric contraction, denoting the contribution of additional factors other than time.

Spirited debate continues over the exact mechanisms through which enhanced work output is facilitated by MTC pre-activation (102). However, experts agree that the SSC has the capacity to enhance the power output of the MTC. Evidence suggests that the primary contributory factor facilitating enhanced performance is the time taken to complete a SSC. Wilson et al. (102) stated, “Generally, it is found that shorter stretch-shorten cycles rely more heavily on the reuse of elastic energy than slow stretch-shorten cycles” (p. 1707). Schmidtbleicher (85) separated the SSC into two classifications based on slow and fast eccentric-concentric contraction times associated with respective movements. These movements were quantified using ground contact times to distinguish between slow and fast SSC. A GCT of less than 0.25 seconds is considered fast, where a GCT greater than 0.25 seconds is considered slow. Slow SSC were also characterized by
large angular displacements at the hip, knee and ankle joints, as well as slower contraction times. Conversely, the fast SSC is exemplified by smaller angular displacements and shorter eccentric-concentric coupling rates. Komi et al. (62) made critical exclusion criteria for an effective fast stretch shorten cycle, which eliminate both the static vertical jump and countermovement jump as valid measures. They asserted 3 essential elements of an efficient SSC including: 1) a well timed pre-activation of the muscle before the eccentric phase, 2) a short eccentric phase, less than 0.25 seconds, and 3) an immediate transition between stretch and shortening.

These criteria led to the use of drop jumps in the present study, rather than SJ or CMJ, because the DJ necessitates significant stretch velocities and rapid transitions from stretch to shortening phases.

**Fascicle vs. Series Elastic Component Contributions to Movement**

The MTC is not entirely efficient at reutilizing stored elastic energy in subsequent work (31, 34, 53). The relative contribution of elastic energy stored in the SEC is of primary interest. Alexander (1) suggested that a specialized structure such as the Achilles tendon is capable of returning as much as 93% of the stored energy during recoil. Evidence suggests that reutilization of stored elastic energy is highly time dependent, with excessive time utilization resulting in energy waste in the form of heat (7). Direct measurement of tendon deformation, through the use of in vivo transducers or optical fibers, has been problematic in terms of human ethics considerations (61). Studies have attempted to quantify tendon deformation using morphological and mechanical data obtained from cadaver external tendons (14, 94, 95). However, the elastic properties of
human cadaver tendons are likely to differ significantly from those of living persons (87).
Modeling studies have also been utilized to compare the contributions of the CE and
elastic components of the MTC in myodynamic performance (4, 17, 18, 94, 95).
Unfortunately, these types of models necessitate information regarding a number of
variables that is not readily available for human muscles (56). However, in recent
research ultrasonography has helped to uncover many of those variables, in order to gain
valuable insight into the role of elastic energy in the SSC (102). Ultrasonic apparatus can
be used in vivo to measure elongation and deformation of the muscle fascicles and
tendinous structures (42).

Kubo et al. (63) used in vivo dynamics to explore the behavior of the human
medial gastrocnemius MTC during two speeds of ankle plantar flexion. They discovered
that tendinous tissue elongated to a greater extent during high velocity plantar flexion.
Notably, in the high speed condition the first fifty percent of plantar flexion produced no
appreciable change in fascicle length. During the final 10° of plantar flexion the entire
MTC showed rapid shortening, creating a catapult effect. These results indicate virtual
isometric contraction of the fascicles throughout the transition from countermovement to
shortening. Similar isometric contraction of the medial gastrocnemius was observed by
Fukunaga et al. (43), during an in vivo study observing walking mechanisms. The stance
phase produced isometric contraction from the muscles while eliciting 7 mm of
elongation from the tendons. The catapult mechanism is again evident in the push-off
phase, as the MTC rapidly shortens. Ishikawa et al. (59) corroborated these results
finding that the stance phase of walking demonstrates slow lengthening of the Achilles
tendon with rapid recoil near termination of ground contact.
The aforementioned rapid recoil, or catapult mechanism, has been hypothesized to increase the SEC external energy release through rapid force decay during shortening (3, 53). However, Ettema (35) showed strong evidence to the contrary, with in vivo examination of rat gastrocnemius muscles through a range of SSCs. Accelerated force decay led to stretching of the CC during shortening, which in several cases showed a decrease in efficiency of the SSC. In fact, researchers have proposed that the use of high levels of elastic energy, stored in the SEC, is not compatible with optimal work production in the CC (10, 36). This discrepancy was explained by citing exorbitant amounts of SEC external energy storage as a cause for limited CC lengthening during stretch and decreased CC shortening during contraction (35). Notably later research by Ettema (34) conflicted with his earlier results. A compliant SEC allowed the CC to generate high work loads. The difference highlighted between these conflicting results centers on the work generated by the CE during eccentric stretch, rather than concentric contraction alone.

The relative contributions of the muscular CC and elastic components to dynamic human performance are still the source of much scholastic disagreement. Controversy centers around three primary factors including: 1) the type of muscle action necessitated by each phase of the movement, 2) preactivity level of muscle, and 3) time to complete movement. Action type refers to terms such as concentric, eccentric, and isometric. Preactivity refers to muscle activation patterns, which can be evinced through electromyography (EMG). Time to complete the movement pattern can be determined through GCT in any movement that has an airborne phase. Otherwise, the initiation of movement can be observed through EMG activity or by using a force plate.
**Stiffness**

A pivotal concept, regarding the amount of muscle and tendon deformation under an external load, is stiffness (21). The term originates in physics and can be defined as “the property of a system to resist an applied stretch” (67, p. 653). Biomechanically, stiffness is used to indicate the linear relationship between force/deformation in bodily materials, including bone, tendon and ligament. It can be calculated by dividing the change in force by the change in length ($\Delta F/\Delta L$) of the relevant material (67). This study will place particular emphasis on muscular stiffness, joint stiffness and leg stiffness.

The stiffness of the musculature surrounding the relevant joints will be responsible for determining the amount of deformation demonstrated by the tendons. Whiting and Zernicke (99) defined muscle stiffness as “activated cross-bridges within the myofibrils that exhibit a resistance to stretching, thus generating an internal force” (p. 46). Extremely compliant muscle is going to waste large amounts of elastic energy internally, but it generates the highest net muscle power and more contractile work (34). In contrast, stiff muscle will generate high peak forces, while exerting large amounts of external work and producing little net power (34). Despite the increased work demands placed on stiff muscle, stiffness may actually produce several mechanical advantages for dynamic muscle performance.

Stiff muscles are thought to act nearly isometrically, during myodynamic activites, leading to increased contractile force (102). The quasi-isometric action avoids the lowered force output that generally accompanies increasing velocity (106). Kurokawa et al. (65) used vertical jumping, without countermovement, to demonstrate that quasi-isometric contraction of the fascicles produced substantial power readings and high joint
angular velocities during late push-off. Researchers have described this type of interaction between muscle and tendinous structures as an energy-saving mechanism (2, 53). The energy-saving mechanism is proposed to operate most efficiently when the MTC shortening is entirely the result of the SEC. In this scenario, the CC is only required to utilize metabolic energy for force generation. Additionally, the quasi-isometric action increases the force per cross-sectional area of active muscle fibers, by controlling sarcomere lengths and ensuring near optimal filament overlap (65).

Stiffness in a single joint can be affected by several factors including muscle activation (98), joint angle (46), angular velocity and range of motion (60). The stiffness of a particular joint can be derived by dividing the change in joint moment by the change in joint angle (ΔM/Δθ) (67). Despite this knowledge regarding contributory factors and calculation methods, the exact mechanisms controlling joint stiffness are not well understood. Co-contraction of opposing agonist and antagonist muscles surrounding the joint is one known method of joint stiffness regulation (76, 77). The nature of multi-joint movements complicates joint stiffness control mechanisms, as energy is exchanged between joints, muscles, tendons and ligaments. This concert of energy exchange is pivotal for efficient movement (30, 48). Much research has focused on the stiffness of the lower extremities, called leg-spring stiffness, using spring-mass models (70, 73).

Wilson and Flanagan (102) defined leg-spring stiffness as “an integration of the stiffness of all lower limb musculoskeletal structures during locomotion describing those structures ability to interact in unison in a spring-like fashion”. The basic spring-mass model is composed of a point mass equivalent of body mass supported by a single linear leg-spring (38). The leg-spring vertical stiffness (K_{vert}) is calculated by dividing the peak
vertical force acting upon the spring \((F_{\text{peak}})\) by the vertical displacement of the center of mass of the spring \((\Delta L)\), at the instant the leg-spring is maximally compressed \((38)\). This model is advantageous in its simplicity, using a single spring to evaluate the mechanical behavior of the entire lower extremity musculoskeletal system. However, much detail is lost in this simplification, including the viscoelastic properties and contributions of the contributing structures \((64)\).

**Contribution of Stiffness to Human Performance**

Research has shown strong correlations between stiffness measures and various indexes of performance such as force output \((103)\), velocity \((64)\), and utilization of elastic energy \((9, 44)\). In the early 1990s, two studies indicated that leg-spring stiffness remained constant during running at different speeds, in both humans and animals \((37, 47)\). Subsequent research has shown strong evidence to the contrary, indicating that leg stiffness increases in relation to velocity \((5, 64)\). These differences have been partially explained through variations in calculation methods \((5)\). Farley et al. \((37)\) concluded that the body increased leg stiffness to accommodate higher running stride frequencies, noting that the vertical displacement decreased as stride frequency increased. Stiffness of the ankle joint, in particular, has proven to be higher in sprinting activities than jogging \((89)\). Arampatzis et al. \((5)\) have argued that increases in running velocity correlate strongly to variations in knee joint stiffness, rather than ankle stiffness. Supporting this proposal, Kuitunen et al. \((64)\) found that knee joint stiffness increased as running velocity increased. In order to effectively utilize elastic energy in myodynamic activities high levels of muscular stiffness are obligatory \((6, 44)\).
Measures of economy, the amount of energy used to perform a certain amount of work, have also shown strong correlations to stiffness. Dalleau et al. (30) used treadmill running to elucidate a negative linear relationship between energy cost and leg stiffness. The research of Dutto and Smith (32) and Heise and Martin (48) highlighted the dramatic effects of fatigue on vertical leg stiffness. An increase aerobic running demands produced a negative effect on vertical leg stiffness. Less economical runners were not able to maintain vertical leg stiffness, adapting a more compliant leg spring mechanism (48). The authors proposed that runners altered their running kinematics to lengthen their stride, in order to maintain velocity while decreasing stride frequency and stiffness (32).

It is evident that stiffness depends not only on the individual and the biological structures involved, but on the demands of the activity (89). Myodynamic activities in particular have demonstrated high correlations between stiffness and decreased GCT. The research of Arampatzis et al. (6) provides critical insight into the relationship between stiffness and GCT, as well as mechanisms controlling stiffness. In this study fifteen participants were asked to perform DJ from three different heights. Experimenters requested that each successive jump to be completed faster, while maintaining efforts to jump as high as possible. Several significant relationships emerged from this data. As stiffness increased GCT decreased, indicating that it may be possible to control leg stiffness by means of GCT. Additionally, attempts to decrease GCT showed higher levels of pre-activation in the leg musculature. Higher muscle pre-activation led to increased stiffness, which produced higher maximal GRF and lower vertical displacement of the center of mass. Nigg and Liu (78) supported these conclusions, regarding muscle pre-activation and vertical stiffness, by studying impact forces during running. Their results
showed that differences in EMG of the pre-activation phase may cause variations in leg stiffness.

Increases in pre-activation of the prime agonist musculature, along with eccentric activity, may lead to significant increases in concentric force output (71). McBride et al. (71) evaluated 3 jumping movements with increasing levels of muscle pre-activity and eccentric load before concentric action, comparing the static jump (SJ), counter-movement jump (CMJ) and drop jump, respectively. The data showed a significant increase in force output throughout the three jumps, escalating with MTC pre-activation and eccentric load. The authors speculated that by increasing pre-load, from CMJ to DJ, the contribution of the eccentric phase to concentric jump performance is enhanced. However, they caution that improvement will only occur when a positive energy balance is maintained throughout the jump. Energy balance becomes negative when the negative work of the eccentric phase is greater than the positive work of the concentric phase. This delicate balance can be affected by the height of the jump and regulation of leg-spring stiffness.

Decreases in GCT are the result of increased leg stiffness, through decreased joint angles, increased pre-activation and decreased electromechanical delay. Shorter GCTs will certainly evince decreased joint flexion, especially in the hip, knee and ankle (64). Electromechanical delay (EMD) occurs when a time lag exists between the increase in electrical activity in the MTC and mechanical response of the muscle (52). The MTC will not transmit any force to the skeletal system until the slack is taken up from the SEC (52). Muraoka et al. (75) revealed that the EMD of the MTC increases if there is slack in the major tendon associated with concentric joint movement, thus increasing GCT and
decreasing the use of elastic energy. For example, by increasing dorsiflexion in the ankle joint, slack will be removed from the Achilles tendon leading to decreased EMD in the gastrocnemius and soleus. Therefore, by decreasing the EMD through stretching the MTC, via muscle pre-activity, GCT is reduced and stiffness is increased. Decreased GCT will lead to amplified returns of elastic energy stored during the contact phase (6, 64, 102).

Pre-activation of the relevant musculature can intensify pre-stretch of the tendinous tissue, which increases the storage of elastic energy (58). Ishikawa et al. (58) observed the behavior of the MTC during maximal depth jumps. They revealed that the fascicles decrease their length change, shifting the majority of elongation to the tendons. Similarly, Sugisaki et al. (90) used RJ to isolate the behavior of the MTC. The tendinous tissues contributed approximately 85% to the mechanical work of the MTC, during the plantar flexion phase. The dynamic jumping research of Arampatzis et al. (6) indicated that an optimum, as opposed to maximum, level of MTC pre-activation may exist in order to maximize mechanical power. They concluded that “maximization of mechanical power is attained through optimum leg stiffness and ankle stiffness values as well as the optimum amount of activation in the muscles of the lower extremities during the pre-activation phase” (p. 363). Therefore, varying levels of leg stiffness and pre-activation are capable of maximizing vertical takeoff velocity and mechanical power.

Stiffness Control Mechanisms, Injury Prevention and Training Applications
At a neural level, the central nervous system (CNS) is ultimately responsible for controlling stiffness. The CNS uses both feedforward and feedback mechanisms to
regulate stiffness (69). The feedforward mechanism consists of the CNS transmitting information directly to the working alpha and gamma motor neurons (84). Gamma motor neurons will innervate the contractile component of the spindle fibers, increasing stretch and tension of their equatorial region. Two primary effects emerge from this innervation: 1) stretch in the spindle fibers depolarizes them enabling direct communication with the alpha motor neurons, increasing local muscle intrinsic activity; 2) it also makes the spindles more excitable to further displacement. This process is thought to aid in reflex activity and fast SSCs (84). The feedback mechanism takes information from the periphery, via proprioceptors, and communicates back to the CNS.

These mechanisms are both evident in Kuitunen et al. (64) who observed increased pre-activation of the plantar flexors and knee extensors, as running velocity increased. In preparation for the impending high impact loads of ground contact, these muscles were pre-activated in order to increase stiffness. Additionally, feedback control was evident in the stretch reflex activity of the soleus muscle, which increased in response to the mechanical loads. Also, the work of Arampatzis et al. (6) is a critical example of feed-forward control. Subjects were able to increase vertical stiffness in response to verbal commands requesting diminished GCTs.

Previously, the negative correlation between fatigue and stiffness was discussed (30, 32, 48). Additionally, stretching prior to exercise appears to have significant performance ramifications. Fowles et al. (41) used 13 maximally tolerable passive stretches to evince a 27% decline in muscular stiffness. The authors proposed that 60% of this initial decrement was attributable to neural factors, including mechanoreceptor activation and Golgi tendon organ inhibitory factors. Multiple studies have shown that
acute static stretching can degrade myodynamic performance (13, 27, 28, 29, 41). A myriad of possible effects have been proposed including: reduced MTC stiffness, decreased muscle activation, reduced force production, decreased reaction time, and reduced performance in CMJ, maximal voluntary contractions and balancing tasks.

Stiffness has also been implicated in both injury prevention and causation. Exorbitant stiffness levels have been associated with dangerous impact forces, increased loading rates contributing to potential musculoskeletal injuries (22, 74). Butler et al. (22) stated, “Increased peak forces, loading rates, and shock can contribute to a greater risk of bony injuries including knee osteoarthritis and stress fractures.” On the other hand, stiffness has also been linked to soft-tissue injury prevention and rehabilitation (22, 82). Stiffness regulation has been characterized as the first line of defense against neuromuscular perturbation (55). Houk (55) described the process through which increased MTC stiffness produces decreased EMD, enabling muscles to neutralize destructive joint forces. Joint stability is therefore enhanced through efficient utilization of stiffness. Stiffness regulation can be described as an inverted U; excessive stiffness may contribute to skeletal injuries while insufficient amounts may contribute to soft-tissue injuries (102).

Research in the field of stiffness training has provided hopeful possibilities for future breakthroughs. Neuromuscular training, including balance and perturbation training have been linked to prevention of ACL injury (57). Plyometric training has been linked to increased energy efficiency (88, 92), improved myodynamic performance and injury prevention (49, 51). Plyometric exercises have been defined as rapid powerful movements that are preceded by a preloading counter-movement that creates a SSC,
including DJ, hopping and bounding (66). The increased stiffness developed through plyometric training can be associated with efficient use of elastic energy through control of the SSC (88). Timing is critical to the SSC. The transition from the eccentric to the concentric phase of movement must be minimized to produce a more powerful action (100). The reactive strength index (RSI) has been developed as a measurement of an individual’s ability to utilize the SSC, changing quickly from eccentric to concentric action (104). RSI is calculated by dividing the height jumped, in a depth jump, by the GCT (72). The index can be utilized to make recommendations for optimal drop jump height, to compare athlete plyometric capabilities and to monitor plyometric training programs (72).
METHODS

Experimental Approach to the Problem

This investigation utilized a repeated-measures design to compare kinetic and kinematic variables of four depth jump conditions. Eighteen healthy junior college athletes performed depth jumps from a 45 cm box under instruction to achieve either maximum jump height or minimal GCT, with either “toes-up” (Figure 1A), dorsiflexed, or “toes-down” (Figure 1B), plantar flexed. These variations in instruction created four distinct jump conditions dorsiflexed-fast (DFF), dorsiflexed-slow (DFS), plantar flexed-fast (PFF) and plantar flexed-slow (PFS), which were drawn at random by the subjects. Biomechanical data was recorded using 3-dimensional motion analysis and force platforms. The four jump conditions were evaluated against the dependent variables GCT, JH, RSI, ankle angle at contact (AAC), peak ankle dorsiflexion angle (PAFA), ankle range (AR), knee angle at contact (KAC), peak knee flexion angle (PKFA), knee range (KR), hip angle at contact (HAC), peak hip flexion angle (PHFA), hip range (HR), initial GRF and peak GRF using a 2x2 analysis of variance (ANOVA).

Subjects

Eighteen male junior college football players voluntarily participated in this study (Table 1). Permission from the college’s athletic director was obtained before the investigation began (Appendix A). All subjects were involved in a year-round strength and conditioning program, which included plyometric jump activity. Before the onset of physical activity, all subjects completed a Modified Par-Q (Appendix B) to eliminate subjects with limitations due to injury or illness. Approval for the use of human subjects
was obtained from this institution’s Standing Advisory Committee on the Protection of Human Subjects. Informed consent documentation (Appendix C) was obtained from each subject before the onset of testing.

**Instrumentation**

Kinematic data was recorded using 7 Vicon MX Cameras (Vicon Motion Systems Ltd., Oxford, UK) at a 240 Hz sampling rate. The cameras were synchronized to 2 Kistler Force Platforms (Amherst, NY, USA) which collected kinetic data at a sampling rate of 1200 Hz, one under each foot. Segments of the lower extremities were identified by a modified Helen Hayes marker placement, positioned on the right and left anterior superior iliac spine, posterior superior iliac spine, thigh, knee, tibia, ankle, toe and heel. A Vertec jump apparatus was used as a secondary measure of jump height. The kinetic and kinematic data was determined using Vicon Nexus software (Version 1.6.1, Vicon Motion Systems Ltd., Oxford, UK).
PROCEDURES

The subjects were instructed to refrain from strenuous physical activity for 2 days prior to testing, including lower extremity weight lifting and plyometrics. A standardized 10 minute warm-up (Appendix D), consisting of both dynamic mobility and low level plyometric phases, along with 10 minutes of jump instruction (Appendix E) were utilized to prepare the subjects for depth jump activity. The dynamic mobility phase involved 10 exercises performed over a 10 yard span that were chosen to properly activate and stretch the lower extremities. The plyometric phase consisted of 2 repetitions of 5 different skipping exercises also performed for 10 yards, along with 2 practice depth jumps in each of the 4 jump conditions.

Subjects randomly drew lots to select the order in which they performed the jump conditions. Each jump trial consisted of a depth jump off of a 45cm box, in which subjects landed with one foot on each of the 2 force platforms and rebounded reaching to touch the fingers of a Vertec measurement device, at the apex of their jump. The jump conditions, DFF, DFS, PFF and PFS, required the participants to step off the box with their “toes-up”, dorsiflexed, or “toes-down”, plantar flexed, and to achieve either minimal GCT or maximal JH. Depth jumps were executed until 3 successful trials of each condition could be recorded. The subjects were given 30 seconds rest between each jump and 1 minute recovery between each jump condition. A certified strength and conditioning professional was present during the testing procedures to limit potential injuries and ensure a safe environment.
Data Analysis

The dependent variables used were GCT, JH, RSI, AAC, PAFA, AR, KAC, PKFA, KR, HAC, PHFA, HR, initial GRF and peak GRF. Two force plates were used for data collection, one under each foot, however, right leg data only was used for data analysis, with the assumption of bilateral symmetry. The force platform data was used to obtain GCT, initial and peak GRF for each jump. Ground contact time for each trial was calculated by multiplying the number of frames the subject was on the ground by .00417. Jump height for each jump was calculated by subtracting the height of the RASIS marker during each subject’s static trial from the pinnacle height of the RASIS reached during each jump and was confirmed with the Vertec results.

The joint angles of the ankle, knee and hip were recorded by the Vicon Nexus 1.61 software. The angle of each joint at the moment of ground contact was subtracted from the peak flexion angle reached at any point during ground contact to calculate joint range. The Reactive Strength Index was calculated by dividing the height jumped by the GCT (104).

Statistical Analysis

All statistical analysis was conducted using PASW for Windows, release 18.0. The four jump conditions were compared using a 2 x 2 (foot x speed) factorial ANOVA with repeated measures ($\alpha \leq .05$). Significant interactions were assessed using post-hoc pairwise comparisons on the four jump conditions ($\alpha \leq .05$).
RESULTS

Ground Contact Time and Jump Height
In regards to GCT, the main effect for speed was significant (p < .001; Table 2). The GCT in the fast conditions was 24.38% faster than the slow conditions. However, the main effect for foot condition was not significant (p = .574). Additionally, no significant interaction was present between the foot and speed conditions (p = .966).

For JH, the main effect for speed (p = .198) and foot (p = .860) condition both were not significant. There was no significant interaction between foot and speed (p = .478: Table 2).

Reactive Strength Index
The RSI showed a significant main effect for speed condition (p < .001: Table 2). The RSI of the fast conditions was 28.76% greater than the RSI of the slow conditions. The main effect for foot was not significant (p = .445) and there was no significant interaction between foot and speed (p = .745).

Ankle Angle and Range
The ankle angle at contact evinced the main effect for foot as significant (p = .017). The dorsiflexed conditions were 28.87% more flexed than the plantar flexed conditions, at the instant of ground contact. Technically, all four conditions landed in plantar flexion. But the PFF and PFS conditions were simply less flexed than the dorsiflexed conditions (Figure 3A). The main effect for speed was not significant (p = .771) and there was no significant interaction between foot and speed (p = .610).
The peak ankle dorsiflexion angle reached, during ground contact, showed a significant main effect for speed (p < .001). The fast conditions displayed achieved 6.64% less dorsiflexion than the slow conditions (Figure 3B). The main effect for foot was not significant (p = .153) and there was no significant between foot and speed (p = .833).

The ankle range, during ground contact, displayed a significant main effect for foot condition (p = .047). The ankle range of the plantar flexed conditions was 8.71% larger than that of the dorsiflexed conditions (Figure 3C). The main effect for speed was not significant (p = .104) and there was no significant interaction between foot and speed (p = .743).

**Knee Angle and Range**

The main effect for both foot (p = .476) and speed (p = .436) were not significant for the knee angle at contact (Figure 3A). Additionally, there was no significant interaction between foot and speed (p = .135).

However, the peak knee flexion angle displayed a significant main effect for speed (p < .001). The fast conditions evinced 10.75% less maximum knee flexion than the slow conditions (Figure 3B). The main effect for foot was not significant (p = .185) and there was no significant interaction between foot and speed (p = .903).

Similarly, the knee range, during ground contact, showed a significant main effect for speed (p < .001). The knee range in the fast conditions was 16.69% smaller than in the slow conditions (Figure 3C). The main effect for foot was not significant (p = .637) and there was no significant interaction between foot and speed (p = .334).
**Hip Angle and Range**

The hip angle at contact indicated a significant interaction between the variables, foot and speed ($p = .041$; Figure 3A). However, post-hoc analysis using pairwise comparisons evinced no significant differences between any of the four variable conditions. Essentially, the HAC interaction between foot and speed was a statistical anomaly. Additionally, the main effect for foot ($p = .754$) and speed ($p = .771$) both proved to be non-significant.

The peak hip flexion angle, during ground contact, showed a significant main effect for speed ($p < .001$). The slow conditions showed 10.84% more hip flexion than the fast conditions (Figure 3B). The main effect for foot was not significant ($p = .799$) and there was no significant interaction between foot and speed ($p = .832$).

The hip range indicated both a significant interaction between foot and speed ($p = .036$; Figure 3C), as well as a significant main effect for speed ($p < .001$). The hip range of the fast conditions was 43.77% smaller than the slow conditions. Post-hoc pairwise comparisons of the four jump conditions revealed significant differences between the fast and slow conditions, as well as the dorsi and plantar foot conditions. The DFF condition was significantly different from DFS ($p < .001$) and from PFS ($p < .001$). The PFF condition was significantly different from DFS ($p < .001$) and from PFS ($p < .001$). However, the fast conditions were not significantly different from each other ($p = .170$) and the slow conditions were not significantly different from one another ($p = .099$). The main effect of the foot condition was not significant ($p = .953$).
Initial and Peak Ground Reaction Force

The initial ground reaction force displayed a significant main effect for foot condition (p = .005) and speed condition (p = .048). The initial GRF for the dorsiflexed conditions was 62.79% greater than the plantar flexed conditions. The fast conditions resulted in 21.74% more initial GRF than the slow conditions. However, there was no significant interaction between the foot and speed conditions (p = .268; Figure 4).

Similarly, the peak ground reaction force also showed a significant main effect for both foot condition (p = .004) and speed condition (p < .001). The peak GRF for the fast conditions was 22.14% greater than the slow conditions. The dorsiflexed conditions resulted in 19.11% greater peak GRF than the plantar flexed conditions. There was no significant interaction between the foot and speed conditions (p = .183; Figure 4).
DISCUSSION

The current investigation demonstrates that decreased ground contact time does not necessitate a significant loss in jump height. In this study, it was hypothesized that variations in foot and speed condition would elicit significant differences in ground contact time and jump height. The fast conditions showed decreased GCT, decreased joint range for the knee and hip, decreased peak flexion at all 3 joints, as well as increased initial and peak GRF. Contrary to our hypotheses, the foot condition was not related to a significant main effect for GCT or JH. However, the foot condition did result in a significant main effect for ankle angle at contact, ankle range, initial GRF and peak GRF. Neither the foot nor the speed conditions displayed a significant main effect for jump height.

Numerous studies have attempted to identify an optimum depth jump height or ground contact time (8, 19, 68, 79, 81, 95, 96). Goss-Sampson et al. (45) proposed 0.26 seconds as the optimal DJ ground contact time. Wilson et al. (101) stated that power production can be significantly reduced if GCT exceeds 0.25 seconds. Additionally, previous studies (19, 97) have found that reduced GCT may lead to reduced vertical velocity. Therefore, our hypothesis stated that shorter GCT would correlate negatively to JH. However, the fast jump conditions, which evinced 24.38% less GCT, did not produce significantly lower jumps heights. These results stand in contrast to the research of Van Ingen Schenau et al. (93) who asserted that muscles could produce more positive work with increased time to build up force development. Bobbert et al. (16) also supported this research by demonstrating a direct correlation between vertical ground reaction force and the time to stimulate the gluteus maximus. It is conceivable that elastic energy return is
responsible for the subject’s ability to reach statistically equivalent jump heights with significantly less GCT. Many researchers have asserted that elastic energy is major contributor to enhanced work and power production, through the SSC (39, 50, 106). This mechanical energy is stored in the SEC of the MTC during the eccentric phase of the SSC can be reused during the concentric phase to enhance contractile performance (52, 90, 93). Wilson et al. (102) asserted that faster SSC rely more heavily on the reuse of elastic energy than slow SSC. The present study displayed significantly faster GCT for the speed conditions. The speed conditions also evinced 22.14% higher peak GRF than the slow conditions. These increased GRF may be the result of increased contribution of the SEC and increased stiffness in the MTC. Previous research has shown that decreased GCT will lead to amplified returns of elastic energy stored during the eccentric phase (6, 64, 102).

The mean ground contact times of the fast conditions, DFF (0.30 s) and PFF (0.31 s), were significantly slower than the 0.25 seconds recommended by Wilson et al. (101) for power production. Our intention was to separate the fast and slow conditions according to the model presented by Schmidtbleicher (85). His research delineated SSC type through GCT defining fast as, less than 250 ms, and slow, as greater than 250 ms. However, in other studies where subjects have been instructed to jump as high as possible without regard for GCT the average GCT were 0.4 seconds (19, 97). Smith et al. (86) also required subjects to perform DJ off of a 45 cm box using 3 conditions: no visible goal, a Vertec measuring device and a hurdle, which subjects has to jump over. The Vertec jump condition resulted in a mean GCT of 0.39 seconds. The hurdle jump condition produced a 0.31 second GCT, which is roughly equivalent to the GCT of the fast conditions in the present study. Young and colleagues (105) demonstrated that jumps utilizing longer GCT
(0.4 s) are an entirely different motor skill than those that require shorter GCT (0.2 s).
The results of this study, as well as Smith et al. (86), seem to fall directly in the middle of those two categories. It is possible that this research asked subjects to execute a task requiring an ideal blend of elastic energy use and motor unit action, stimulated by the use of an external goal. This study required not only that subjects get off the ground as fast as possible, in the speed conditions, but also that they attempt to reach touch a maximal Vertec marker, on every trial. The research of Smith et al. (86) also reported increased peak vertical velocity and vertical GRF with decreased GCT. These results evince possible relationship between increased peak vertical velocity, GRF, decreased GCT and the use a goal, as an additional stimulus.

In the present study, the speed conditions resulted in 21.74% greater initial GRF and 22.14% more peak GRF than the slow conditions. Walsh et al. (97) conducted drop jump research in which decreased GCT correlated with increased maximal vertical GRF. The Smith et al. (86) Vertec condition explained 51% of the increased vertical GRF through decreased GCT. For the speed conditions to attain statistically equivalent jump heights to the slow conditions it was necessary for the subjects to generate more GRF impulse, in less time. The RSI results support this conclusion as the fast conditions displayed 28.76% more reactive strength than the slow conditions. The RSI is designed to elucidate the ability to generate maximal force in a minimal amount of time (107, pg 28). The current investigation demonstrates that subjects are able to decrease their GCT and achieve similar JH through reactive strength, by increasing their rate of force development and increasing the amount of force generated. A key component contributing to reduced GCT is stiffness of the MTC (6).
The speed conditions showed significantly less joint flexion than the slow conditions, with 6.64% less ankle, 10.75% less knee and 10.84% less hip flexion respectively. Additionally, the speed conditions displayed significantly smaller joint ranges at the knee, 16.69%, and hip, 43.77%. The drastic decrease in joint range and maximum joint angle are clear indicators of increased stiffness. These results suggest that subjects may manipulate joint angle, during ground contact, through muscular stiffness, in an effort to get off the ground faster. Arampatzis et al. (6) performed DJ research in which subjects were required to control GCT. Results indicated that subjects were able to increase MTC stiffness in order to decrease GCT. Additionally, the study suggested that higher muscle pre-activation contributed to the increase in joint stiffness. The current investigation hypothesized that the dorsiflexed foot conditions would increase muscle pre-activation and decrease electromechanical delay, thus, decreasing ground contact time.

The results, however, showed no significant difference between GCT attributable to foot condition. Lack of EMG data makes it impossible to quantify muscle pre-activation in this study. However, the foot condition did have a significant main effect on both initial and peak GRF, as well as ankle angle at contact and ankle range. Subjects landed 28.87% more dorsiflexed in the DFF and DFS conditions than in the PFF and PFS conditions, producing 62.79% more initial GRF. Additionally, the ankle range of the dorsiflexed conditions were 8.71% less than the plantar flexed conditions, while the peak GRF was 19.11% greater. The load on the lower extremity musculature was clearly higher at ground contact for the dorsiflexed conditions. McBride et al. (71) proposed that increased pre-activation and eccentric load of the MTC would enhance jump
performance. However, a point of diminishing returns has been proposed whereby the eccentric load can be too great and subsequent concentric contraction exhibits no further increase in force production or elastic energy return (103). Wilson et al. (103) proposed that this loss of contractile ability is the result of a delayed amortization phase, eccentric to concentric shift. This delay can lead to the dissipation of elastic energy as heat, thus reducing power production (45, 54). The load can be altered through points of contact, speed, drop height, weight of participant or external load (80). Peng, (79) suggested that drop jumps above 40 cm correlated with an incremental decrease in leg and ankle stiffness, as well as an increase in GCT. Farley et al. (38) found that subjects exhibited extended limb posture, decreased knee flexion and increased ankle plantar flexion during drop jumps from 40 cm and these biomechanical alterations are amplified by increasing the height to 60 cm. It is possible that the 45 cm drop jump height, used in the present study, increased the eccentric load contributing to the lack of a significant main effect for foot condition, in regards to both GCT and JH. However, reported values for optimum jump heights include 0.4 – 0.62 meters (8, 81). Ball et al. (12) recommended 0.6 m drop height to ensure no bilateral differences in vertical GRF and temporal components. In this study, drop heights of 0.2 m and 0.4 m contributed to laterality, or significant bilateral differences in vertical GRF. These lower box heights also produced shorter GCT then the 0.6 m condition. However, the 0.6 m condition showed significantly decreased time to peak force from first contact. Research suggests that decreases in time to peak GRF are indicative of increased lower leg stiffness with increasing drop height (11, 83). Future research may want to examine time to peak GRF, as well as evaluating the current four jump conditions using different box heights.
In conclusion, our findings suggest that depth jump GCT can be reduced without a significant corresponding decrease in jump height. A ground contact time of 0.3 s may be the minimal GCT possible to elicit maximum jump height. The use of a goal, like a Vertec, may generate greater jump heights, knee flexor moments and effort (40). The variation of foot condition, through dorsiflexion, may not affect GCT or JH, but it does increase initial and peak GRF. Increased GRF may be desired if the goal is to attain increased reactive strength and rate of force development. By decreasing maximum joint angles at the ankle, knee and hip during ground contact through increased muscular stiffness GCT can be reduced and GRF may increase.
PRACTICAL APPLICATIONS

This study has confirmed previous research that asserted GCT can be altered by subject manipulation of joint stiffness at the knee, ankle and hip. Decreased GCT does not necessitate a drop off in jump height, if GRF is increased and elastic energy is utilized through the SSC. The depth jump is an effective exercise to practice controlling stiffness through GCT. The use of an external goal or barrier may lead to increased jump performance and shorter GCT. The optimal GCT for maximum depth jump height may be close to 0.3 s, which is slower than the 0.26 s optimal GCT reported by Goss-Sampson et al. (45). However, depth jumps over 40 cm may be too great to achieve fast stretch-shorten cycles, less than 250 ms. Increased dorsiflexion of the ankle at ground contact does not significantly affect ground contact time or jump height, thus, it is not recommended at this time. Future research should revisit these four jump conditions at depth jump heights above and below 45 cm. Additionally, the longitudinal effects of depth jump training should be examined.
REFERENCES


7. Aruin, AS and Prilutskii, BI. Relationship of biomechanical properties of muscles to their ability to utilize elastic deformation energy. Human Physiol 11: 8-12, 1985.


APPENDIX A

LA Pierce Permission Letter

California State University, Northridge
Standing Advisory Committee for the Protection of Human Subjects
18111 Nordhoff Street
Northridge, CA 91330-8232

Dear Committee Members:

   Joshua Phillips has permission to conduct the project entitled “Relationship between ankle joint angle and ground contact time in athlete drop jump performance” at the Los Angeles Pierce College, Woodland Hills, California. I have reviewed the project and am aware of all of the activities involved in the project including: student medical clearance, modified Par-Q questionnaire, jump training, depth jump testing and sprint testing.

Signed,

Date:
APPENDIX B

Modified Par-Q

Start by answering the ten questions in the box below. If you are between the ages of 15 and 69, the PAR-Q will tell you if you should check with your doctor or a member of the LAPC medical staff before you start. Common sense is your best guide when you answer these questions. Please read the questions carefully and answer each one honestly: write YES or NO.

1. Has your doctor ever said that you have a heart condition **and** that you should only do physical activity recommended by a doctor?
2. Do you feel pain in your chest when you do physical activity?
3. In the past month, have you had chest pain when you were not doing physical activity?
4. Do you lose your balance because of dizziness or do you ever lose consciousness?
5. Do you have a bone or joint problem (for example, back, knee or hip) that could be made worse by a change in your physical activity?
6. Is your doctor currently prescribing drugs (for example, water pills) for your blood pressure or heart condition?
7. Do you know of **any other reason** why you should not do physical activity?
8. Have you experienced fever, sore throat or vomiting within the last week?
9. Has your diet or sleep schedule changed significantly in the last week?
10. Do you have a minor injury that you have not yet reported to the UCSB medical staff?

**YES to one or more questions (Questions 1-7)**
Talk with your doctor by phone or in person BEFORE participating in this study. Tell your doctor about the PAR-Q and which questions you answered YES.

**YES to one or more questions (Questions 8-10)**
Talk with a member of the LAPC training staff BEFORE participating in this study. Tell the training staff which questions you answered YES.

**NO to all questions**
If you answered NO honestly to all PAR-Q questions, you can be reasonably sure that you can participate in this study.
APPENDIX C

Participant Informed Consent

CALIFORNIA STATE UNIVERSITY, NORTHRIDGE

Project Title: Relationship Between Ankle Joint Angle and Ground Contact Time in Athlete Drop Jump Performance

Introduction:
This study, conducted by Joshua Phillips and Sean Flanagan, PhD, is designed to compare the efficiency of four distinct jumping techniques.

Description of Research:
This research will add to our understanding of how ankle angle effects ground contact time and jumping performance. The results will contribute to future instruction of jumping technique and may lead to increased performance and injury prevention.

You will be asked to travel to the California State Northridge Kinesiology biomechanics lab to test depth jump performance using four distinct conditions. After Informed Consent is given, you will complete a Modified Par Q (see Appendix 1) form asking about your current health status. Various measurements will be taken, including height, weight and reach. Sixteen retro-reflective markers will be placed on key landmarks on both sides of your body to identify joint positions. Subsequently, you will complete a 10 minute dynamic warm-up (see Appendix 3) and 10 minutes of jump instruction (see Appendix 2). Next you will perform depth jump trials at a predetermined height, and rebound off of a force platform, reaching for the highest possible Vertec marker. You will execute four jumping techniques selected in random order including: Dorsiflexed fast (DFF), Dorsiflexed slow (DFS), Plantar-Flexed fast (PFF) and Plantar-flexed slow (PFS). Subsequently, you will proceed to the track where two maximal velocity 40 meter sprint trials will be timed using two laser timing tripods. You will be given at least 30 seconds of rest between each jump and 1 minute between each jumping condition and sprint trial. You may request additional rest time between trials and it will be freely granted. You must perform each depth jump and sprint trial with maximal effort. You may stop your participation at any time. You can expect to be in the lab at least an hour.

Subject Information and Risks:
Risks are minimal as jumping and sprinting are a regular part of your athletic training. Risks include fatigue, muscle soreness, strains and sprains. Before you are accepted for participation in the study completion of a physical exam and full medical clearance will be confirmed by the Los Angeles Pierce College athletic training staff or you will complete a Modified Par-Q (see Appendix 1). You will complete a standardized dynamic warm-up (see Appendix 3) and standardized jump training (see Appendix 2) to familiarize yourself with the testing protocols.
The standardized jump training and the standardized warm-up will both be progressive in nature. The standardized warm-up will begin with dynamic flexibility exercises and progress to moderate plyometric exercise. The standardized jump training will build from lower eccentric loads to higher eccentric loads. You may experience irritation or an allergic reaction to the tape. If an injury were to occur on the CSUN campus, you will be referred to the health center, a local physician, or their primary care physician all at your own cost.
You may withdraw from the study at any time, without consequence.

Confidentiality & Final Disposition of Data:
All information collected during this project will be held in strict confidence as required by law. Confidentiality will be maintained by using alphanumeric codes rather than personal identifying information. The data and copies of the consent form will be kept in a locked cabinet at the investigator’s office. The study results may be published in scientific journals or presented at scientific meetings but the names or identity of participants will not be made known. Upon publication of the results, any data identifying the participants will be destroyed.

Benefit of Participation:
You will not receive monetary compensation for participation in this study. However, the results may enable us to make recommendations regarding the plyometric training, which may contribute to improved athletic performance.

Concerns:
If you wish to voice concern about the research, you may direct your questions to Research and Sponsored Projects, 18111 Nordhoff Street, California State University, Northridge, Northridge, CA, 91330-8232, or phone 818-677-2901. If you have specific questions about the study you may contact Dr. Sean Flanagan at 18111 Nordhoff Street, Northridge, CA 91330-8287 or by phone at (818)677-7507. You may also contact Joshua Phillips by phone at (310)691-0456.

Voluntary Participation:
You should understand that your approval to participate in this study is completely voluntary, and you may decline to participate or withdraw from the study at any time without jeopardy. Likewise the researcher may cancel this study at any time. A copy of your informed consent will be stored in a locked filing cabinet and you will be given a copy of this informed consent for your records.

Videotaping:
During the course of the project participants will be videotaped. Your initials ______ here signify your consent to be videotaped. All tapes collected as part of this project will be kept on file by the researcher at the conclusion of the study.
I have read the above and understand the conditions for participation in the described study. I understand that in no way does signing this form remove any of my legal rights nor does it relieve the investigators, sponsors or involved institutions from their legal and professional duties. I give consent to participate in the study.

Participant’s Name _______________________________________________
(please print) Last First MI

Street Address ___________________________________________________

City _____________________ State _________ Zip _________

**Subject Signature:**

Signature ______________________________ Date ______________

**Witness Signature:**

Witness/P.I. Signature __________________________ Date ______________

If you have signed this form, please return one copy in an envelope to:
Dr. Sean Flanagan
Department of Kinesiology
California State University, Northridge
18111 Nordhoff St
Northridge, CA 91330-8287
APPENDIX D

Dynamic Warm-up

All exercises will cover 10 yards

**Dynamic flexibility:**
Heel Walk, Toe Slap Walk, Front Lunge, Reverse Lunge with Overhead Reach, Alternating Side Lunge, Inchworm, SL Reverse RDL, Quad Stretch, Hip Flexor Raise, World’s Greatest Stretch

**Plyo’s:**
Ankle Hops x 2, Speed Skips x 2
Drop Skip x2, Cross Over Skip x 2
High Knee Skip x2
APPENDIX E

Jump Training Protocol

Two Footed Jump w/dorsiflexion (2 sets 30 seconds)

Ankle Mobility Exercise (2 sets, 3 directions, 5 reps each)

4 Jump Conditions (5 jumps each condition): Dorsiflexed fast (DFF), Dorsiflexed slow (DFS), Plantar flexed fast (PFF) and Plantar flexed slow (PFS)
APPENDIX F

Table 1. Physical characteristics of the subjects (mean ± SD)

<table>
<thead>
<tr>
<th>Age (y) (kg)</th>
<th>Height (cm)</th>
<th>Mass</th>
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</thead>
<tbody>
<tr>
<td>21.01 ± 1.34</td>
<td>181.31 ± 6.97</td>
<td>89.99 ± 14.47</td>
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</tbody>
</table>

Table 2. GCT, jump height and RSI variables.*†

<table>
<thead>
<tr>
<th>Variable</th>
<th>DF</th>
<th>DS</th>
<th>PF</th>
<th>PS</th>
</tr>
</thead>
<tbody>
<tr>
<td>GCT (sec)</td>
<td>0.3026 (0.0536)‡</td>
<td>0.4007 (0.0782)¶</td>
<td>0.3055 (0.0579)‡</td>
<td>0.4041 (0.0779)¶</td>
</tr>
<tr>
<td>JH (cm)</td>
<td>58.13 (11.58)</td>
<td>59.25 (11.16)</td>
<td>58.33 (10.48)</td>
<td>58.82 (9.78)</td>
</tr>
<tr>
<td>RSI (mm/ms)</td>
<td>1.995 (0.546)‡</td>
<td>1.563 (0.509)¶</td>
<td>1.987 (0.539)‡</td>
<td>1.53 (0.461)¶</td>
</tr>
</tbody>
</table>

*GCT = ground contact time; RSI = reactive strength index; DF = dorsifast condition; DS = dorsislow condition; PF = plantarfast condition; PS = plantarslow condition.

†Data is presented as mean (SD).

‡Significantly different from DS & PS conditions (p < .001)

¶Significantly different from DF & PF conditions (p < .001)
Figure 1. (A) Dorsiflexed “toes-up” condition; (B) Plantarflexed “toes-down” condition.
Flexion Angle at Contact (deg)

Ankle
Knee
Hip

Peak Flexion Angle (deg)

Ankle
Knee
Hip
Figure 2. Joint angles during ground contact. (A) Joint flexion at the instant of foot contact; (B) peak joint flexion during ground contact phase; (C) joint angular range during ground contact phase. \( ^{a} \) Significantly different from slow conditions DS, PS; \( ^{b} \) Significantly different from fast conditions DF, PF; \( ^{c} \) Significantly different from “toes-down” conditions PF, PS; \( ^{d} \) Significantly different from “toes-up” conditions DF, DS; \( ^{e} \) Significant interaction between foot and speed conditions. Differences were considered significant when \( p < .05 \).
Figure 3. Initial ground reaction force (IGRF) and peak ground reaction force (PGRF) during ground contact phase. a Significantly different from slow conditions DS, PS; b Significantly different from fast conditions DF, PF; c Significantly different from “toes-down” conditions PF, PS; d Significantly different from “toes-up” conditions DF, DS. Differences were considered significant when p < .05.