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Research article

Joint Kinetics to Assess the Influence of the Racket on a Tennis Player's Shoulder

Thomas Creveaux [1,2], Raphaël Dumas [1,3], Christophe Hautier [1,2], Pierre Macé [4], Laurence Chèze [1,3] and Isabelle Rogowski [1,2]✉

[1] Université de Lyon, Lyon, France; [2] Centre de Recherche et d'Innovation sur le Sport, Villeurbanne Cedex, France; [3] Laboratoire de Biomécanique et Mécanique des Chocs, Villeurbanne Cedex, France; [4] Babolat VS, Lyon, France

Abstract

This study aimed at investigating the influence of three rackets on shoulder net joint moments, power and muscle activity during the flat tennis serve under field-conditions. A 6-camera Eagle® motion analysis system, operating at 256 Hz, captured racket and dominant upper limb kinematics of the serve in five tennis players under three racket conditions (A: low mass, high balance and polar moment, B: low three moments of inertia, and C: high mass, swingweight and twistweight). The electromyographic activity of six trunk and arm muscles was simultaneously recorded. Shoulder net joint moments and power were computed by 3D inverse dynamics. The results showed that greater shoulder joint power and internal/external rotation peak moments were found to accelerate and decelerate racket A in comparison with the racket C. Moreover, serving with the racket A resulted in less activity in latissimus dorsi muscle during the acceleration phase, and biceps brachii muscle during the follow-through phase when compared with racket C. These initial findings encourage studying the biomechanical measurements to quantify the loads on the body during play in order to reduce them, and then prevent shoulder injuries. Racket specifications may be a critical point for coaches who train players suffering from shoulder pain and chronic upper limb injuries should be considered in relation to the racket specifications of the players.

Key words: EMG, inverse dynamics, joint power, joint moment, tennis serve.

Introduction

Major research in tennis racket innovation is based on analytical models that mimic impacts on the racket-face, in order to understand the influence of racket specifications on the player's performance (Allen et al., 2011). However, especially during the manufacturing design process, the effects of manipulating a racket, with optimal specifications defined under simulated or laboratory conditions, are usually based on the player's subjective evaluation (Statham, 2007). Even if the player's feeling is an important criterion in the racket choice, this feeling has a low correlation with the variations in the racket specifications (Statham, 2007). Focusing on objective measurements could help to highlight the player's body loads as they relate to racket handling during tennis movements.

Previous studies have mainly focused on the relationships between the racket specifications and racket power and manoeuvrability. Based on simulated data, serve speed after the impact is increased when the racket mass increases or when the balance point moves towards the tip (Haake et al., 2007). However, the change in mass and/or balance point results also in a change of swingweight (Cross, 2001). In addition, a racket with a small moment of inertia is easy to swing but gives less power, i.e. less ball rebound speed, than a racket with a larger moment of inertia, for a given racket head speed (Brody, 2000). During an internal rotational movement of the upper limb carried out at maximal effort, the swing speed remains approximately constant as the mass of a rod increases when keeping the moment of inertia fixed, while the swing speed increases as only the moment of inertia of the rod decreases (Cross and Bower, 2006). Under real serve conditions, a decrease in racket moment of inertia can significantly increase the racket head speed, suggesting that, in the player's hand, a lighter racket would be easier to accelerate, and hence more powerful than a heavier one (Mitchell et al., 2000). However, all these previous works did not focus on the loads on the player's body when playing with powerful or easy to use rackets.

Objectivization of player-racket interaction under field conditions is mainly based on vibration or electromyographic outcomes. Vibration analysis detects no effect of string vibration dampers on racket frame vibration transfer to the forearm (Li et al., 2004). Muscle activity analysis reveals no influence of grip size on forearm muscle patterns (Hatch et al., 2002), while the shoulder muscle activity is inversely related to the racket mass (Rogowski et al., 2009). In addition, Marx et al. (2001) report that 'the lever arm and weight of the racket markedly increase stress on the shoulder and the potential for injury'. These previous findings suggest that it would be easier to detect the player's adaptation to racket specifications at the proximal joint, and that joint moment and power outcomes remain an unexplored mechanism in the study of racket effects on joint kinetics during tennis strokes. Considering the potential for injury related to tennis practice, the study of the influence of racket specifications on both performance and biomechanical requirements at the shoulder joint appears as a critical concern for tennis players and coaches.

A study investigating the influence of three different rackets on the upper limb muscle activity and the shoulder net joint moments and power during the flat tennis serve under field conditions was conducted. The first hypothesis is that different rackets result in measurable differences of muscle activity, joint moments and powers during the tennis serve. The second hypothesis is that, for similar post-impact ball speed, serving when using a racket with low moment of inertia would result in high moment of inertia.

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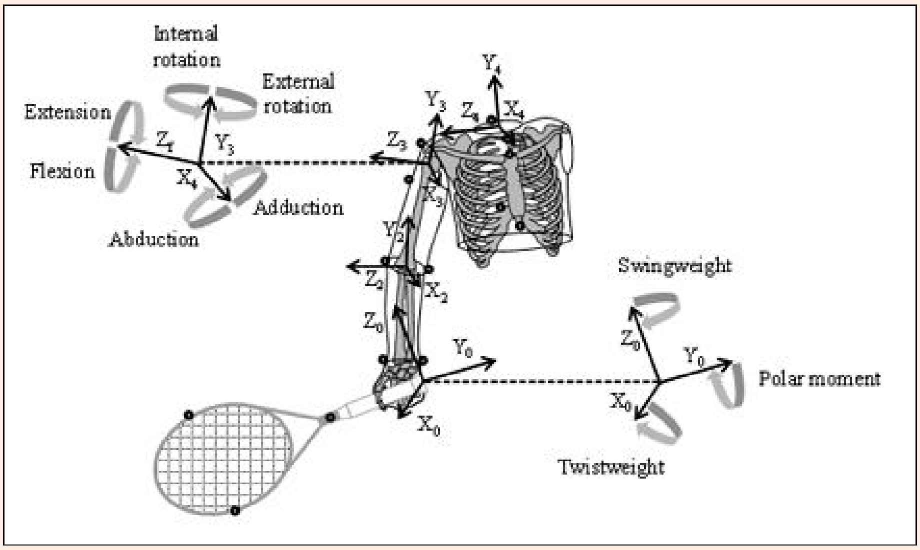

Figure 1. Location of the markers (o) on the trunk, dominant upper limb and racket of the tennis player. Black arrows indicate the segment coordinate system (X_i, Y_i, Z_i) with i=0, the racket/hand segment; i=2, the forearm segment; i=3, the upper arm segment, and i=4, the thorax. The axes X_4 and Y_3, as well as a floating axis Z_f define the joint coordinate system and the movement orientations of the shoulder. The pivoting axis to measure the swingweight and twistweight was fixed at 10 cm from the handle extremity (*).

Methods

Participants

Five male players (Mean ± Standard deviation: age = 25 ± 4 years; height = 1.81 ± 0.07 m; mass = 77 ± 9 kg; skill = International Tennis Number 3) volunteered to participate in this study. Players gave their written informed consent and this study was approved by the ethical committee “Sud-Est II”.

Rackets

Three tennis rackets, noted A, B and C, were tested. The specifications of these rackets were measured after the rackets were strung (250 N) and grip tape applied, as well as after the video markers were pasted on the racket-frame. The mass, balance point (defined by the distance, in mm, from the racket shaft handle to the location of the centre of mass) and the moments of inertia, defined according to Figure 1, were determined using the Babolat Racquet Diagnostic Center® (Babolat, Lyon, France) for the balance point, swingweight and twistweight, and using the Babolat laboratory oscillation test for the polar moment (Figure 1). As reported in Table 1, racket A presented low mass, high balance and polar moment; racket B was characterized by low moments of inertia, and racket C by high mass and moments of inertia.

Experimental design

Performance was measured on an indoor acrylic tennis court. After warming-up, each participant performed, randomly, eight flat serves for each racket with 3-min intervals between rackets. The player was instructed to hit flat serves, i.e. with minimal spin, in the deuce diagonal with similar ball velocity for the three rackets; after impact, the ball had to bounce in a target zone defined in the external corner of the serve box, one metre inside from the sideline and one metre inside from the serve line. A radar gun (SR3600, Sports-radar, Homosassa, FL, USA) was placed behind the player to measure the ball velocity after the racket/ball impact, and give feedback on ball velocity to the player.

Videographic recordings

Fifteen spherical reflective markers were attached on the subject and racket to define the coordinate systems of the thorax, upper arm, forearm, and racket. The markers were attached to the xiphoid process, incisura jugularis (suprasternal notch), C7, T8, and, on the dominant side, angulus acromialis, deltoïdus tuberosity, medial and lateral epicondyles of the elbow, radial and ulnar styloid processes (Wu et al., 2005). On the nondominant side, the markers were placed on angulus acromialis and radial styloid process. Two markers were attached at mid-height

Table 1. Characteristics of the three regular (R) and video-instrumented (VI) rackets (strung and with grip tape). The pivoting axis used to measure the swingweight and twistweight was fixed at 10 cm from the handle extremity, according to the axis system (X_0, Y_0, Z_0) defined in the Figure 1.

	A		B		C	
	R	VI	R	VI	R	VI
Mass (g)	302	340	320	359.8	365	387.4
Balance (mm)	321	331	302	321	320	326
Swingweight (kg·cm²)	292	346	278	331	342	368
Twistweight (kg·cm²)	338	364	320	346	356	385
Polar moment (kg·cm²)	13.9	17	11	14.4	13.6	16.4

of both racket-face sides to determine the centre of the racket-face and one marker was placed at the top of the handle (Figure 1). Retro-reflective tapes were placed around the ball to detect the ball–racket impact. The six-camera Eagle® motion analysis system (Motion Analysis Corp., Santa Rosa, CA, USA) collected the 3D trajectories of markers during serves at a sampling rate of 256 Hz.

To quantify the movement of the dominant upper limb, this study modelled a four-segmented linkage system, including the thorax, upper arm, forearm and hand-racket (Figure 1), and assumed each body segment was a rigid body. All the trajectories of the reflective markers were smoothed using a triangular filter kernel obtained from two passes of a 20 points sliding average window (Campione and Gentilucci, 2011). Segment coordinate systems (SCS) of each upper limb segment were constructed according to International Society of Biomechanics recommendations (Wu et al., 2005). The gleno-humeral joint centre was estimated by regression (Reed et al., 1999; Dumas et al., 2007). Upper limb net joint moments ($\mathbf{F}_i$, $\mathbf{M}_i$) were computed by a 3D inverse dynamic method (Dumas et al., 2004; Cleather and Bull, 2010). Inputs for the computation were the quaternion attitude and the origin of each segment i ($\mathbf{q}_i$ and $\mathbf{r}_{Pi}$), the mass m_i, position of centre of mass $\mathbf{r}_{Ci}{}^s$ and matrix of inertia $\mathbf{I}_i{}^s$ (at the centre of mass) of body segments estimated by regression in the SCS (Dumas et al., 2007) and of the racket (i=0), computed at the racket centre of mass from the Table 1 values using parallel axis theorem:

$$\begin{bmatrix}\mathbf{F}_i\\ \mathbf{M}_i\end{bmatrix}=\begin{bmatrix}\mathbf{E}_{3\times3} & \mathbf{0}_{3\times3}\\ \widetilde{\left(\mathbf{r}_{C_i}-\mathbf{r}_{P_i}\right)} & \mathbf{E}_{3\times3}\end{bmatrix}\begin{bmatrix}m_i\left(\mathbf{a}_i-\mathbf{g}\right)\\ \mathbf{I}_i\boldsymbol{\alpha}_i+\tilde{\boldsymbol{\omega}}_i\mathbf{I}_i\boldsymbol{\omega}_i\end{bmatrix}+\begin{bmatrix}\mathbf{E}_{3\times3} & \mathbf{0}_{3\times3}\\ \widetilde{\left(\mathbf{r}_{P_{i-1}}-\mathbf{r}_{P_i}\right)} & \mathbf{E}_{3\times3}\end{bmatrix}\begin{bmatrix}\mathbf{F}_{i-1}\\ \mathbf{M}_{i-1}\end{bmatrix}$$

where $\boldsymbol{E}_{3x3}$ is the identity matrix, $\boldsymbol{0}_{3x3}$ a zero matrix, $\mathbf{g}$ the acceleration of gravity, and ~ is the skew matrix.

The position of centre of mass ($\mathbf{r}_{Ci}$) in the inertial coordinate system (ICS), the matrix of inertia ($\mathbf{I}_i$) in the ICS, the linear acceleration of the centre of mass $\mathbf{a}_i$, the angular velocity and acceleration of the segment (ω_i and α_i) were all computed with the quaternion algebra (i.e., $\otimes$ is the quaternions product and $^{-}$ the quaternion conjugate (Dumas et al., 2004)):

$$\begin{bmatrix}0\\ \mathbf{r}_{C_i}\end{bmatrix}=\mathbf{q}_i\otimes\begin{bmatrix}0\\ \mathbf{r}_{C_i}^s\end{bmatrix}\otimes\bar{\mathbf{q}}_i$$

$$\mathbf{I}_i=\mathbf{R}_i\mathbf{I}_i^s\mathbf{R}_i^T \text{ with } \mathbf{R}_i=\left[\left(q_{i0}{}^2+\mathbf{q}_{iv}{}^T\mathbf{q}_{iv}\right)\mathbf{E}_{3\times3}+2\mathbf{q}_{iv}\mathbf{q}_{iv}{}^T+2q_{i0}\tilde{\mathbf{q}}_{iv}\right]$$

$$\text{where } \mathbf{q}_i=\begin{bmatrix}q_{i0}\\ \mathbf{q}_{iv}\end{bmatrix}$$

$$\begin{bmatrix}0\\ \mathbf{a}_i\end{bmatrix}=\begin{bmatrix}0\\ \ddot{\mathbf{r}}_{P_i}\end{bmatrix}+\ddot{\mathbf{q}}_i\otimes\begin{bmatrix}0\\ \mathbf{r}_{C_i}^s\end{bmatrix}\otimes\bar{\mathbf{q}}_i+2\left(\dot{\mathbf{q}}_i\otimes\begin{bmatrix}0\\ \mathbf{r}_{C_i}^s\end{bmatrix}\otimes\dot{\bar{\mathbf{q}}}_i\right)+\mathbf{q}_i\otimes\begin{bmatrix}0\\ \mathbf{r}_{C_i}^s\end{bmatrix}\otimes\ddot{\bar{\mathbf{q}}}_i$$

$$\begin{bmatrix}0\\ \boldsymbol{\omega}_i\end{bmatrix}=2\dot{\mathbf{q}}_i\otimes\bar{\mathbf{q}}_i \text{, and } \begin{bmatrix}0\\ \boldsymbol{\alpha}_i\end{bmatrix}=2\left(\ddot{\mathbf{q}}_i\otimes\bar{\mathbf{q}}_i+\dot{\mathbf{q}}_i\otimes\dot{\bar{\mathbf{q}}}_i\right)$$

Each time derivative of quaternion required for the inverse dynamics (i.e., in $\mathbf{a}_i$, ω_i, and α_i) was followed by filtering (4th order Butterworth filter with 12 Hz cut frequency). The recursive computation starts for the wrist net joint forces moments with the segment (i=1*) composed of the hand (i = 1) and the racket (i = 0). The hand centre of mass is supposed fixed in the racket axes (at -4 cm along Y_0 axis, Figure 2). As no markers were placed on the hand, the quaternion attitude and the origin of the hand-racket SCS are:

$$\mathbf{q}_{1*}=\mathbf{q}_0 \text{ and } \mathbf{r}_{P_{1*}}=\mathbf{r}_{P_0}$$

The mass m_{1*}, position of centre of mass $\mathbf{r}_{C1*}{}^s$, and matrix of inertia $\mathbf{I}_{1*}{}^s$ (at the centre of mass) in the hand-racket SCS are:

$$m_{1*}=m_1+m_0$$

$$\mathbf{r}_{C_{1*}}^s=\frac{m_1\mathbf{r}_{C_1}^s+m_0\mathbf{r}_{C_0}^s}{m_{1*}} \text{ and}$$

$$\begin{aligned}\mathbf{I}_{1*}^s=\ & m_1\left(\left(\left(\mathbf{r}_{C_1}^s-\mathbf{r}_{C_{1*}}^s\right)^T\left(\mathbf{r}_{C_1}^s-\mathbf{r}_{C_{1*}}^s\right)\right)\mathbf{E}_{3\times3}-\left(\mathbf{r}_{C_1}^s-\mathbf{r}_{C_{1*}}^s\right)\left(\mathbf{r}_{C_1}^s-\mathbf{r}_{C_{1*}}^s\right)^T\right)\\ &+\mathbf{I}_0^s+m_0\left(\left(\left(\mathbf{r}_{C_0}^s-\mathbf{r}_{C_{1*}}^s\right)^T\left(\mathbf{r}_{C_0}^s-\mathbf{r}_{C_{1*}}^s\right)\right)\mathbf{E}_{3\times3}-\left(\mathbf{r}_{C_0}^s-\mathbf{r}_{C_{1*}}^s\right)\left(\mathbf{r}_{C_0}^s-\mathbf{r}_{C_{1*}}^s\right)^T\right)\end{aligned}$$

Outputs of the computation were the net 3D joint moments ($\mathbf{M}_3$) acting at the dominant shoulder joint in the ICS. Shoulder joint power (P_3) was computed as dot products of 3D joint moments and the difference between distal and proximal segment angular velocities:

$$P_3=\mathbf{M}_3\cdot\left(\boldsymbol{\omega}_4-\boldsymbol{\omega}_3\right)$$

The 3D shoulder joint moment were then expressed in the joint coordinate systems (JCS) (Desroches et al., 2010; Morrow et al., 2009):

$$\mathbf{M}_3=\underbrace{\frac{\left(\mathbf{e}_2\times\mathbf{e}_3\right)\bullet\mathbf{M}_3}{\left(\mathbf{e}_1\times\mathbf{e}_2\right)\bullet\mathbf{e}_3}}_{M_{e1}}\mathbf{e}_1+\underbrace{\frac{\left(\mathbf{e}_3\times\mathbf{e}_1\right)\bullet\mathbf{M}_3}{\left(\mathbf{e}_1\times\mathbf{e}_2\right)\bullet\mathbf{e}_3}}_{M_{e2}}\mathbf{e}_2+\underbrace{\frac{\left(\mathbf{e}_1\times\mathbf{e}_2\right)\bullet\mathbf{M}_3}{\left(\mathbf{e}_1\times\mathbf{e}_2\right)\bullet\mathbf{e}_3}}_{M_{e3}}\mathbf{e}_3$$

with ($\mathbf{e}_1$, $\mathbf{e}_2$, $\mathbf{e}_3$) the three axes of the JCS.

In order to avoid gimbal lock during tennis serve, these axes were the $\mathbf{X}_4$ axis of thorax, $\mathbf{Z}$ floating and $\mathbf{Y}_3$ of upper arm (Bonnefoy-Mazure et al., 2010). Positive moments were shoulder adduction, flexion, and internal rotation. The joint angles were computed using the same JCS and with the same positive/negative conventions. The functional significance of the joint moment patterns in terms of energy generation and storage/dissipation can be understood by examining the positive and negative powers, respectively. The frame of impact is not presented (vertical black line in the Figure 3) because the results of inverse dynamics ($\mathbf{M}_3$, P_3) are not interpretable without having measured, or estimated (Haake et al., 2003; Yang et al., 2012), the impact forces. However, the methods of impact force estimation have not been used for inverse dynamics under field conditions so far. The frame of impact in the present study was automatically identified as the time where the absolute normal distance of the ball to the racket plane is minimal (and simultaneously the absolute tangential distance of the ball to the centre of the racket face is inferior to the face radius). Both the smoothing of markers trajectories (triangular filter kernel) and the filtering of quaternions (4th order Butterworth filter) were performed on the raw data including frame of impact. Only the results of inverse dynamics at the frame of impact were discarded, since they cannot be interpreted.

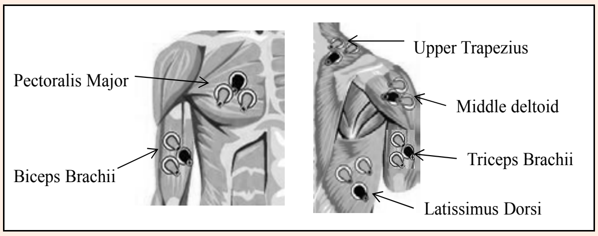

Figure 2. EMG electrodes location.

Electromyographic recordings

To assess the EMG activity of selected muscles, seven surface electrodes (EMG Triode Electrode, Nickel-plated brass, interelectrode distance = 2 cm, Thought Technology, Montreal, Canada) were attached to the upper trapezius (UT), pectoralis major (PM), latissimus dorsi (LD), middle deltoid (MD), biceps brachii (BB) and triceps brachii (TB) muscles of the dominant side (Figure 2). Skin surfaces were shaved and treated with alcohol before the electrodes were attached. The surface electrodes were located in accordance with the SENIAM recommendations (Surface EMG for Non-Invasive Assessment of Muscles (Hermens et al., 2000)). The EMG signals were collected using the Flexcomp Infiniti system (Thought Technology, Montreal, Canada, 2048 Hz). The EMG signals were analyzed in the time domain, i.e. the raw EMG signals were filtered using a 4^{th} order Butterworth (band pass 10-500 Hz, then the Root Mean Square values (EMGrms, windows: 50 ms) were computed (Rogowski et al., 2009). The EMGrms values were averaged during each phase of the tennis serve (defined from the joint kinematics).

Data analysis

Three phases of the tennis serve were considered during the present study. The downward acceleration phase is initiated from maximal shoulder external rotation and lasts until the minimal height of the racket-face; the upward acceleration from the minimal height of the racket-face to the frame prior impact (-0.004 s); and the early follow-through corresponding to the first 25% duration of the follow-through phase, defined from the frame after impact (+0.004 s) to the completion of the stroke (Morris et al., 1989). Among all serves performed in all racket conditions, only those for which the ball rebound was in the predetermined target zone were considered successful and taken into consideration in the subsequent statistical analysis. Values for ball velocity, durations of the three phases, mean EMGrms for each muscle, minimal and maximal joint moment peaks, and mean negative and positive shoulder joint power during each phase of the serve for the three rackets are presented as mean ± standard error (SE) over the participants. A Friedman test was used to test for significant differences across racket conditions. If significant differences were observed, Wilcoxon rank tests were performed to test for significant differences in each parameter between A and B, A and C, and B and C rackets. All the statistical tests were performed using software SPSS 11.0.1. (Chicago, IL, USA) and level of significance was set at $p \leq 0.05$.

Results

Ball velocities after impact were 36.4 ± 2.3 $m \cdot s^{-1}$ for the racket A, 37.1 ± 1.5 $m \cdot s^{-1}$ for B and 37.2 ± 2.3 $m \cdot s^{-1}$ for C. No significant differences were observed between the three racket conditions. The duration of the three phases presented no significant difference between the three rackets. On average, the duration was 0.23 ± 0.03 s for downward acceleration, 0.14 ± 0.01 s for upward acceleration, and 0.27 ± 0.01 s for the follow-through of the serve.

Figure 3 shows the time histories of moments and power calculated for each of the three rackets in one player. Mean shoulder peak moments and joint power are reported in Table 2. Shoulder adduction, extension and external rotation peak moments presented low values during the first phase of the tennis serve, and then increased strongly during the last two phases. Abduction peak moment displayed a similar pattern, except that its values increased only during the follow-through phase. Flexion and extension peak moment values increased from the downward acceleration phase to the upward acceleration phase, and decreased during the follow-through phase. A significantly higher internal rotation moment was observed for racket A when compared with C during the upward acceleration phase ($p = 0.04$). A significantly higher external rotation peak moment was found during the follow-through phase for the racket A in comparison with racket B ($p = 0.02$) and C ($p = 0.04$). All other comparisons between the three rackets for peak moments revealed no significant differences. In addition, a significantly higher mean positive power during the upward acceleration, as well as a significantly greater mean negative power during the follow-through phase were observed for racket A in comparison with racket C ($p = 0.04$ for both). No other significant differences were observed for the shoulder joint power.

Table 3 presents mean muscle activations expressed in reference to mean muscle activation measured during the downward acceleration phase when serving with racket A. The PM, LD and TB muscle activities increased from the downward to upward acceleration phases, and finally decreased during the follow-through phase. The BB, UD and MD muscle activity increased throughout the upward acceleration and follow-through phases. No significant differences were reported for all muscles between the three racket conditions during the downward acceleration phase. During the upward acceleration phase, the LD muscle activity was significantly lower when serving with racket A in comparison with rackets B (p = 0.03) and C (p =0.02). Finally, during the follow-through phase, only the BB muscle activity was significantly lower when serving with racket A compared with racket C (p = 0.02).

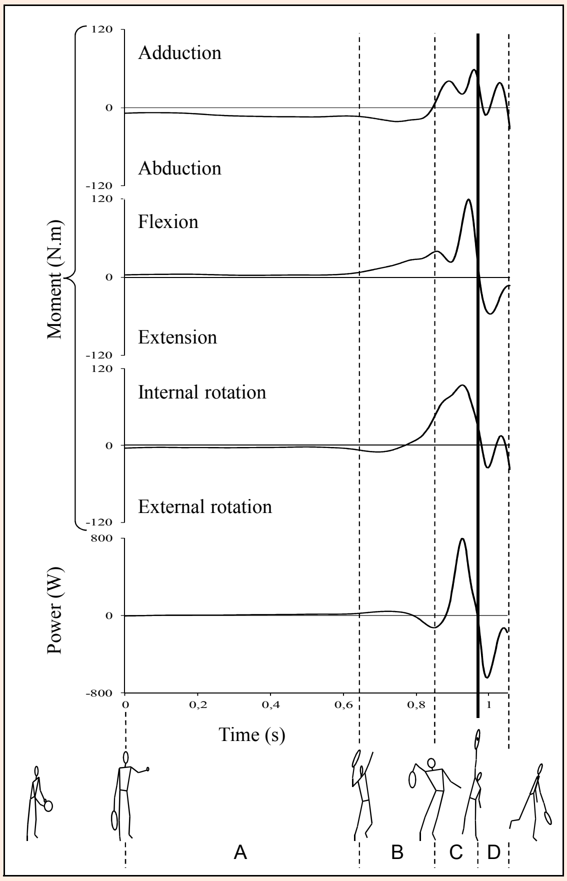

Figure 3. Time histories of shoulder moments and power obtained for a single player with the rackets A (plain black), B (dashed black) and C (plain gray). Vertical dotted lines delimit the four studied phases of the serve, i.e. cocking, downward acceleration, upward acceleration and early follow-through. The plain black vertical line indicates the frame of impact, for which the results of inverse dynamics were discarded. The overlapping of the curves was achieved by normalizing the durations of each phase for the three rackets.

Discussion

The present study investigated the effects of the three different rackets on shoulder joint kinetics during tennis serves. The results suggest that the acceleration and deceleration of racket A required greater power and shoulder internal/external rotation peak moments, as well as less muscle activity than racket C.

The results of the present study showed that shoulder kinetics and muscle activity were moderate to high during the follow-through phase and high during the acceleration phase, in particular during the upward trajectory of the racket face. These findings were consistent with the data reported in the literature (Elliott et al., 2003; Morris et al., 1989; Reid et al., 2007; Ryu et al., 1988; Seeley et al., 2008). Overall, the joint moment values were lower than those reported in the literature (Elliott et al., 2003; Reid et al., 2007). These differences could be explained by the serve instructions given to the players; e.g. maximal effort was expected in previous studies, while similar ball velocity under the three racket conditions was instructed in this study, hence resulting in lower ball velocities (43 $m \cdot s^{-1}$ (Elliott et al., 2003) *vs.* about 37 $m.s^{-1}$ in the present study). The lack of significant differences in ball velocities after the ball/racket impact showed that all players complied with the experimental instructions, irrespective of the racket used. In addition, the three-phase durations of the serve were consistent with the data reported previously in the literature (Seeley et al., 2008), and no significant differences were observed between the three rackets conditions. Consequently, similar ball velocities and temporal pattern of movement allowed the differences in measured values of moments, powers, and EMG activity to be related to the variations in racket specifications.

Before racket/ball impact, the shoulder kinetics and muscle activity measured under the three racket conditions were similar. However, during the upward acceleration phase, racket A (lowest mass, highest values for balance and polar moment; Table 1) needed more joint power than racket C (highest mass, swingweight and twistweight; Table 1) to internally rotate the upper limb (Ryu et al., 1988), and induced a significantly higher internal rotation peak moment (Table 2). It could be hypothesized that a higher velocity may be generated at impact when serving with a light racket in comparison with serving with a heavy one. Indeed the ball velocity after impact is related to the kinetic energy transfer from the racket to the ball, which depends on both racket mass and velocity squared at impact. Serving with a light racket resulted in higher internal rotation moments at the shoulder that would develop an imbalance of internal and external rotators of the shoulder. Such anterior/posterior imbalance has already been highlighted in tennis players (Kibler et al., 1996), and appears as a risk factor for injuries, such as impingement of the shoulder girdle. In addition, six muscles can be recruited to internally rotate the upper arm (pectoralis major, latissimus dorsi, anterior deltoid, subscapularis, teres major and supraspinatus). As reported in Table 3, similar activity between racket conditions was observed for the pectoralis major and anterior deltoid muscles, while significantly lower activity was found for the latissimus dorsi muscle for racket A when compared with racket C (Table 3). It could be hypothesized that the decrease in the latissimus dorsi muscle activity would not contribute enough to resist humeral distraction during overhead movement and could be compensated by a higher activity of some rotator cuff muscles (not evaluated here).

After racket/ball impact, i.e. during the follow-through phase, the critical role of the deceleration phase is to help dissipate the kinetic energy that was not imparted to the ball. Racket A seemed to be more difficult to decelerate than racket C, as greater energy absorption and greater external rotation peak moment were observed (Table 2).

Table 2. Mean (± Standard Error) shoulder peak moments (Nm) and joint power (W).

	Downward acceleration			Upward acceleration			Follow-through		
	A	B	C	A	B	C	A	B	C
Maximal moment peak									
Flexion	26.8 (4.6)	27.2 (3.8)	27.2 (4.8)	58.2 (6.8)	60.0 (6.6	54.5 (7.8)	28.2 (7.3)	33.4 (3.7)	30.5 (7.7)
Adduction	-1.8 (1.7)	-3.1 (1.7)	-3.0 (1.1)	54.8 (13.0)	52.3 (8.8)	53.8 (16.3)	35.9 (4.8)	45.5 (3.1)	42.3 (9.1)
Internal rotation	21.7 (4.4)	21.6 (3.5)	21.6 (4.6)	50.6 † (6.4)	44.8 (5.1)	45.6 (7.7)	18.0 (4.1)	23.5 (3.6)	22.4 (5.5)
Minimal torque peak									
Extension	4.0 (3.6)	4.4 (3.0)	3.3 (3.5)	-75.5 (14.3)	-76.6 (10.0)	-73.4 (19.8)	-64.3 (13.0)	-59.1 (9.2)	-56.4 (16.6)
Abduction	-15.6 (1.9)	-16.5 (1.6)	-16.4 (1.8)	-9.0 (1.1)	-13.5 (.6)	-11.0 (1.6)	-31.5 (6.3)	-35.3 (8.2)	-29.1 (9.0)
External rotation	-2.1 (3.6)	-1.0 (3.6)	-1.6 (3.6)	-15.5 (4.0)	-16.5 (4.8)	-14.8 (4.1)	-25.5 *† (5.5)	-20.5 (4.6)	-21.9 (6.2)
Mean Positive Power	28.0 (8.7)	23.6 (6.8)	26.6 (7.8)	224.0 † (38.3)	196.6 (19.1)	179.2 (35.9)	161.7 (55.9)	203.6 (38.1)	169.7 (49.0)
Mean Negative Power	-7.3 (3.8)	-15.2 (10.4)	-7.4 (4.5)	-276.8 (76.4)	-260.4 (99.4)	-223.5 (79.1)	-279.3 † (77.9)	-223.2 (71.3)	-226.2 (72.3)

* Post hoc comparison between A and B rackets and † Post hoc comparison between A and C rackets, with $p \leq 0.05$.

These greater values may be explained by a higher velocity at impact of the racket A compared with C, as hypothesized previously. Simultaneous lower biceps brachial muscle activity to slow down racket A when compared with racket C (Table 3) suggested that the shoulder external rotator muscles (infraspinatus, teres minor and posterior deltoid) would be more activated to slow down the racket/upper limb complex. A light tennis racket such as racket A would generate fatigue of the rotator cuff muscles during the upward acceleration and follow-through phases, thus reducing the dynamic stability of the shoulder, and therefore increasing the potential risk of shoulder injuries.

Apart from classical issues in motion analysis and the limitation of the number of participants resulting in small statistical magnitude, the three rackets tested in this study displayed interrelated specifications. The physics of the tennis racket implies that its customization by additional mass corresponds to alterations in moments of inertia and balance (Cross, 2001), hence challenging the control of the range of variations in each racket specification. Consequently, the relationship between each racket specification and shoulder joint loadings remains difficult to interpret with certainty. However, this study showed that kinetics outcomes at the shoulder may be interesting indicators to compare tennis rackets under field-conditions. Moreover, the present findings agree with previous results obtained for tennis forehand drive (Rogowski et al., 2009) but may appear as contradictory with previous statement stipulating that a heavy racket increases shoulder stress (Marx et al., 2001).

Conclusion

The results of this study suggested that kinetic strategies were different depending on the racket used to reach similar post-impact ball velocity. The combination of some physical specifications may contribute to increase shoulder joint power and peak joint moments based on unascertained muscular patterns. According to the findings of this study, using a light racket with low polar moment may contribute to increased biomechanical requirements at the player's shoulder joint. Inversely, using a heavier racket with high twistweight and swingweight would result in lower solicitations of the shoulder to achieve a similar

Table 3. Mean (±Standard Error) muscle activation (%) in reference with muscle activation for racket A (100%) during the downward acceleration phase.

	Downward acceleration			Upward acceleration			Follow-through		
	A	B	C	A	B	C	A	B	C
Pectoralis Major	100.00	132.4 (130.3)	153.7 (87.6)	144.3 (63.8)	116.3 (28.6)	146.9 (110.2)	62.3 (55.7)	92.6 (53.5)	93.5 (47.7)
Latissimus Dorsi	100.00	149.4 (150.8)	148.9 (90.5)	205.7*† (180.7)	207.6 (154.8)	250.3 (192.8)	78.8 (26.7)	155.0 (143.2)	102.5 (55.9)
Upper Trapezius	100.00	250.4 (497.2)	293.2 (474.5)	247.3 (178.3)	265.8 (176.6)	296.1 (232.3)	508.6 (549.4)	587.8 (610.5)	612.7 (562.1)
Middle deltoid	100.00	128.7 (214.9)	147.3 (210.0)	453.9 (481.7)	469.2 (451.5)	462.0 (572.2)	556.7 (560.4)	669.7 (619.1)	680.9 (684.1)
Biceps Brachii	100.00	132.8 (228.0)	159.6 (172.8)	429.5 (564.5)	288.5 (188.6)	600.1 (637.0)	391.8† (322.6)	587.6 (287.6)	640.6 (439.4)
Triceps Brachii	100.00	79.8 (35.3)	78.1 (28.3)	222.9 (154.1)	244.7 (188.5)	228.5 (165.6)	113.1 (85.3)	178.1 (218.8)	98.8 (69.9)

* Post hoc comparison between A and B rackets and † Post hoc comparison between A and C rackets, with $p \leq 0.05$
Standard Error were not reported for reference values

performance. If further studies are needed to determine the exact effect of each racket specification on joint loadings, these initial findings encourage biomechanical measurements for quantifying loads on the body during play in order to reduce them, and so prevent shoulder injuries. Coaches should consider carefully the choice of a racket in relation with its specifications, especially when the racket is manipulated by a player suffering from shoulder pains. Furthermore, chronic upper limb injuries should encourage coaches to check for potentially inappropriate racket specifications of their players.

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Key points

- Light racket required more joint power than heavy one to achieve similar post impact ball velocity.
- Serving with a light racket resulted in higher shoulder internal and external rotation moments than using a heavy one for similar performance.
- Chronic shoulder pain should encourage coaches to check for potentially inappropriate racket specifications of their players.

AUTHORS BIOGRAPHY

Thomas CREVEAUX
Employment
Research Engineer at the University Claude Bernard Lyon 1
Degree
PhD
Research interests
Human movement modeling, motor control, signal analysis
E-mail: thomas.creveaux@univ-lyon1.fr

Raphaël DUMAS
Employment
Associate Professor at the University Claude Bernard Lyon 1
Degree
PhD
Research interests
Personalized three-dimensional modeling (i.e., multi-body systems) of the human muscular-skeletal system for applications in orthopaedics, posture and locomotion.
E-mail: raphael.dumas@univ-lyon1.fr

Christophe HAUTIER
Employment
Assistant professor at the University Claude Bernard Lyon 1
Degree
PhD
Research interests
Muscle, ageing, diabetes, obesity, sport ergonomics
E-mail: christophe.hautier@univ-lyon1.fr

Pierre MACÉ
Employment
Member of the Babolat research and development department (Lyon, France)
Degree
Engineer
Research interests
Racket and tennis material development
E-mail: pmace@babolat.com

Laurence CHÈZE
Employment
Professor at the University Claude Bernard Lyon 1
Degree
Engineer, PhD
Research interests
Multi-body modeling of the musculo-skeletal system, 3D motion experimental measurement and analysis for both fundamental research and clinical investigations
E-mail: laurence.cheze@univ-lyon1.fr

Isabelle ROGOWSKI
Employment
Assistant professor at the University Claude Bernard Lyon 1
Degree
PhD
Research interests
Sport engineering and motor performance
E-mail: isabelle.rogowski@univ-lyon1.fr

✉ **Isabelle Rogowski**
UCB Lyon 1 – UFRSTAPS – CRIS EA 647, 27-29, bd du 11 novembre 1918, 69622 Villeurbanne Cedex, France