A robotic apparatus that dictates torque fields around joints without affecting inherent joint dynamics

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ABSTRACT

This manuscript describes how motor behaviour researchers who are not at the same time expert roboticists may implement an experimental apparatus, which has the ability to dictate torque fields around a single joint on one limb or single joints on multiple limbs without otherwise interfering with the inherent dynamics of those joints. Such an apparatus expands the exploratory potential of the researcher wherever experimental distinction of factors may necessitate independent control of torque fields around multiple limbs, or the shaping of torque fields of a given joint independently of its plane of motion, or its directional phase within that plane. The apparatus utilizes torque motors. The challenge with torque motors is that they impose added inertia on limbs and thus attenuate joint dynamics. We eliminated this attenuation by establishing an accurate mathematical model of the robotic device using the Box–Jenkins method, and cancelling out its dynamics by employing the inverse of the model as a compensating controller. A direct measure of the remnant inertial torque as experienced by the hand during a 50 s period of wrist oscillations that increased gradually in frequency from 1.0 to 3.8 Hz confirmed that the removal of the inertial effect of the motor was effectively complete.

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1. Introduction

In coordination studies, torque fields acting about joints are one of the factors which influence observed behaviour. Yet, it is normally difficult to manipulate torque fields without also varying other important factors, such as plane of motion, posture, or the perceptual qualities of the produced pattern of coordination. It is also difficult to manipulate torque fields for one limb without also changing them for other limbs. As a result, separation of the effects of these various factors, and discovery of the true causes of certain observations may remain beyond reach. An instrument which allows the setting of torque fields around limbs independently of one another, independently of posture and of the plane of motion, and without interfering with the inherent dynamics of joints would therefore be a useful experimental aid in the study of coordination.

For example, we have recently used such an apparatus for the right wrist, to explore the familiar experience of tapping in time with a favourite tune, during which the downward phase of the gesture invariably coincides with the beat of the music (Carson, Oytam, & Riek, 2009). We were able to demonstrate that the propensity to move down on the beat arises not because of the much hypothesized perception and cognitive internalization of the direction of terrestrial gravity as an ecological invariant, but simply as a result of its immediate inertial effect on the stability and economy of action. In half the trials, we used the apparatus to create a net torque around the wrist which was equal in magnitude and opposite in direction to the gravitational torque. In other words, in these trials we created an artificial gravity that had the same magnitude as terrestrial gravity, except it pulled up rather than down. If the tendency to move down on the beat stemmed from a structural internalization of the orientation of terrestrial gravity, then temporarily altering the net torque about the wrist should have no or minimal effect on coordination. If, on the other hand, it stemmed from the downward movement being assisted by the net torque, then the reversal of the net torque should significantly influence the stability of synchronization. In its strongest sense, the latter hypothesis would predict that synchronization would reverse (to moving up on the beat) with the reversal of the net torque. In those trials where we used the apparatus to match the magnitude of terrestrial gravity but reverse its direction, it was the upward phase of the gesture that coincided with the beat for all participants without exception.

In this manuscript, we describe the apparatus that enables torque fields around a single joint per limb to be specified without otherwise interfering with the dynamics of that joint. For a device of its kind, it is based on relatively accessible control theory concepts some of which are familiar to the research community, and it does not require the user to compute typically complex equations of motion. The apparatus utilizes an electric torque motor attached to an interface – e.g., a handle, or a pedal depending on the limb. For simplicity, we will refer to the interface as the “handle”. The primary technical problem is that while it is natural to use a torque motor to generate torque, the means of coupling the limb – i.e., a motor-handle system – is far from not interfering with the dynamics of the limb. In fact, if left uncompensated, the coupling imposes added inertia on the limb and attenuates the joint dynamics. This is due primarily to the inherent characteristics of the torque motor, but also to the mass of the handle. This attenuation is particularly troublesome in interlimb coordination tasks, where high frequency oscillations of the limb are typically of particular interest. The higher the movement frequency, the more pronounced the attenuating effect of the motor-handle system. The technical challenge is to control the motor in such a way that we maintain its torque generating capacity while eliminating its attenuating effect on the inherent joint dynamics. In order to manipulate single joints on multiple limbs in parallel, the methods described here are repeated on multiple motor-handle systems.

Our toolset for the task includes a desktop computer equipped with Matlab/Simulink (Mathworks, Natick, MA, USA) software package coupled to a real-time controller board (dSPACE, Paderborn, Germany) which controls the AC servo-motor (Baldor BSM 4250AA, Fort Smith, AR, USA). The sampling rate of the controller board is 200 Hz. The basic function of the motor is that it takes an input signal in volts, \( v_c(t) \), and produces torque in Nm which is equal to \( v_c(t) \) multiplied by a scalar \( (1/k_{Nm\text{to}V}) \) which we know in advance. Essentially, we control the motor by determining \( v_c(t) \). The control of the motor is formulated in a high-level graphical language in Simulink. The Real Time Workshop feature of Matlab translates the Simulink design into machine code, which gets downloaded onto the
controller board. As measurements, we have at our disposal the angular displacement and true angular velocity of the handle, obtained directly from the motor’s encoder, and the reactive torque at the handle derived from a torque sensor mounted between the motor shaft and the base of the handle. We also record not only $v(t)$, the input signal to the motor, but all of the individual components that make it up as part of the control strategy.

2. Method

2.1. Torque motor

Fig. 1 illustrates our apparatus which allows the manipulation of torque-position relation about the wrist by using a torque motor. The hand is placed in a handle attached to the motor. While it is wise to make the handle out of light-weight material (e.g., aluminium), it will nevertheless not be weightless. This is an undesirable addition to the weight of the hand and will contribute to the gravitational torque about the joint. The net torque acting upon the wrist is equal to the gravitational torque ($\tau_g$), plus that which is generated by the motor ($\tau_m$):

$$\tau_{net} = \tau_g(\text{hand}) + \tau_g(\text{handle}) + \tau_m$$

The equations for $\tau_g(\text{hand})$ and $\tau_g(\text{handle})$ can be derived from equations, torque = force $\times$ distance, and force = mass $\times$ acceleration:

$$\tau_g(\text{hand}) = m_{\text{hand}} \cdot g \cdot d \cdot \sin(\theta)$$

where $g$ is the gravitational acceleration and $d$ is the distance between the axis of rotation aligning with the wrist and the centre of mass of the hand, or

$$\tau_g(\text{hand}) = k_{\text{hand}} \cdot \sin(\theta), \quad \text{where} \quad k_{\text{hand}} = m_{\text{hand}} \cdot g \cdot d$$

Similarly,

$$\tau_g(\text{handle}) = m_{\text{handle}} \cdot g \cdot d \cdot \sin(\theta),$$

$$\tau_g(\text{handle}) = k_{\text{handle}} \cdot \sin(\theta), \quad \text{where} \quad k_{\text{handle}} = m_{\text{handle}} \cdot g \cdot d$$

Both $k_{\text{hand}}$ and $k_{\text{handle}}$ are established directly and accurately – as opposed to relying on the theoretical values or estimates of $g$ and $d$. This is done by taking direct measurements of the gravitational torque (by means of the reactive torque sensor) and the angular position of the handle, and then using the relationship, $k = \tau_g / \sin(\theta)$. For $k_{\text{handle}}$, measurements are taken of the handle alone. For $k_{\text{hand}}$, mea-

Fig. 1. The picture on the left depicts our apparatus which enables the manipulation of torque-position relation about the wrist. In the corresponding illustration on the right, $\tau_g$ and $\tau_m$ are torques acting upon the wrist due to gravity and the motor, respectively. $\theta$ is the angular displacement as measured from the vertical.
surements are taken with the hand placed in the handle, which effectively gives us \( k_{\text{hand} + \text{handle}} \). We get \( k_{\text{hand}} \) by subtracting \( k_{\text{handle}} \) from \( k_{\text{hand} + \text{handle}} \). To ensure complete relaxation of the hand (i.e., effectively no muscular activity), amplified audio feedback of the activity of the principal wrist extensor and flexor muscles – extensor carpi radialis (ECR) and flexor carpi radialis (FCR) – is presented to the participants.

Given (1), a desired net torque (\( \tau_{\text{net}} \)) can be achieved by manipulating the motor such that,

\[
\tau_{\text{m}} = -\tau_{g(\text{hand})} - \tau_{g(\text{handle})} + \tau_{\text{net}}
\]

As an illustration, let us suppose that we are interested in “reversing” the direction of gravity as experienced by the limb while maintaining its magnitude, i.e., \( \tau_{\text{net}} = -\tau_{g(\text{hand})} \). In other words, “reverse gravity” has the same magnitude as terrestrial gravity, except it pulls up rather than down. Substituting \( \tau_{\text{net}} = -\tau_{g(\text{hand})} \), (2) and (3) into (4), we get

\[
\begin{align*}
\tau_{\text{m}} &= -\tau_{g(\text{hand})} - \tau_{g(\text{handle})} + \tau_{\text{net}} \\
\tau_{\text{m}} &= -2\tau_{g(\text{hand})} - \tau_{g(\text{handle})} \\
\tau_{\text{m}} &= -(2k_{\text{hand}} + k_{\text{handle}}) \sin(\theta)
\end{align*}
\]

Setting \( \tau_{\text{m}} \) in accordance with (5), we generate the necessary torque for the reversal of gravity about the wrist. It is important to think of \( \tau_{\text{m}} \) as consisting of two components – the compensatory component, \( k_{\text{handle}} \sin(\theta) \) which compensates for the weight of the handle, and the phenomenal component, experienced by the hand, which in this case is \( -2k_{\text{hand}} \sin(\theta) \).

It is worth acknowledging that the supposition underlying the physical model of the hand above and the use of the handle to impose a motor generated torque field is that the axis of rotation of the wrist remains fixed. While we made every effort to fit the participant’s hand in the handle in such a way that the axes of rotation of the wrist and the handle are aligned, it is possible that the axis of rotation of the wrist may change slightly with motion. This is a limitation common to all studies that utilize a handle to either impose a certain torque field or merely to record joint dynamics.

2.2. Inertial (low-pass filter) effect of the motor-handle system

While the employment of the torque motor solves the gravity reversal problem, unfortunately it creates another of its own. The inertia of the motor-handle system attenuates like a low-pass filter the joint dynamics of the limb upon which it operates (Fig. 2). In other words, the limb needs to produce additional torque in the face of this attenuation in order to maintain a particular \( \theta(\cdot) \). We can solve this problem by making the motor produce the extra torque to counteract the inertia of the motor-handle system. With the appropriate input, we can manipulate the motor such that it produces the torque, \( \tau(\cdot) \), which is precisely what is needed to support \( \theta(\cdot) \), the dynamics at the handle being generated by the limb. The limb then would not have to generate the extra torque, \( \tau(\cdot) \), needed to overcome the inertia of the motor in order to maintain \( \theta(\cdot) \) – the motor would do that itself. From the perspective of the limb, therefore, motor inertia would be eliminated. A horse and carriage analogy inspired by Gogol (1842/2004) may be useful here. The carriage weighs on the horse and affects its course of travel, as the handle does on the limb. However, if we were to motorize the carriage and drive it in synchrony with the horse, the horse would no longer be affected. Like Chichikov’s “crafty

![Fig. 2](image-url) The torque motor produces the torque, \( \tau_{\text{m}} \), required to reverse gravity about the wrist when \( K = -(2k_{\text{hand}} + k_{\text{handle}}) \), \( \theta \) is the angular displacement of the handle, \( k_{\text{motor}} \) is the motor specific scaling factor (which translates the unit of the signal from Nm to volts) such that the desired torque fed as input to the motor is produced at the output. However, this setup introduces a problem of its own. The inertia of the motor-handle system acts as a low-pass filter upon the limb, attenuating its dynamics.
dappled horse", it would run along in front of the carriage without pulling it (Gogol, 1842/2004, p. 42). The crucial factor now is to determine the appropriate input to the motor which will make this happen.

2.3. Shaping the input to the motor so as to eliminate low-pass filtering

With a torque motor, what we control readily and directly is the torque it produces – \( \tau(t) \) in Nm is equal to the input signal \( \nu_s(t) \) in volts multiplied by a scalar \( \frac{1}{k_{NmtoV}} \) which we know in advance, \( \tau(t) = \nu_s(t) \times \left( \frac{1}{k_{NmtoV}} \right) \). As such, we can dictate a particular torque \( \tau(t) \) at the handle, but not angular displacement \( \theta(t) \). What we can do, however, is measure \( \theta(t) \) which results from \( \tau(t) \). We can also calculate the mathematical relationship between \( \tau(t) \) and \( \theta(t) \), which we call \( \mathbf{N} \). One way to think of \( \mathbf{N} \) is that it maps \( \tau(t) \) onto \( \theta(t) \); \( \theta(t) = \mathbf{N} \times \tau(t) \). The inverse of \( \mathbf{N} \), that is, \( \mathbf{N}^{-1} \) describes the reverse of this relationship and maps \( \theta(t) \) onto \( \tau(t) \).

Here, we have the beginning of the end of our motor inertia problem. We can realize \( \mathbf{N}^{-1} \) as a real-time filter, as part of the computer control of the motor. When we sample and feed \( \theta(t) \) through this filter and apply digital-to-analogue conversion, it gives us \( \tau(t) \). Multiplying \( \tau(t) \) by \( k_{NmtoV} \), we get \( \nu_s(t) \). Going back to our original problem, when \( \nu_s(t) \) is fed to the motor as input, it generates \( \tau(t) \), the extra torque needed to support \( \theta(t) \) and thus eliminate the inertia of the motor as experienced by the limb.

We have at our disposal from the motor’s encoder, not just angular displacement \( \theta(t) \), but also true angular velocity, \( \dot{\theta}(t) \). With no loss of generality and for reasons of practical advantage, we calculate \( \mathbf{M} \), the mapping between \( \tau(t) \) and \( \dot{\theta}(t) \), and we use \( \mathbf{M}^{-1} \) to generate \( \tau(t) \) and \( \nu_s(t) \). For a torque motor, \( \mathbf{M} \) is simpler than \( \mathbf{N} \), and we know the relationship between them – i.e., \( \mathbf{M} \) is most likely to be a first order low-pass filter, and \( \mathbf{N} \) is equal to \( \mathbf{M} \) multiplied by an integrator. On account of it not having an integrator, system identification techniques identify \( \mathbf{M} \) with more accuracy than they would \( \mathbf{N} \). Furthermore, \( \mathbf{M}^{-1} \) is better behaved as a controller than \( \mathbf{N}^{-1} \) as it does not incorporate noisy differentiation, or differencing as its discrete-time counterpart.

As depicted in Fig. 3, it is wise and in most cases necessary to scale down the contribution of \( \mathbf{M}^{-1} \) to the motor’s input by feeding it through a compensation gain \( k_c \), in order to guarantee that this contribution at no point in time exceeds what is required to overcome the inertia of the motor. The cost of overcompensating for inertia far outweighs the cost of undercompensating. If we undercompensate

![Fig. 3. \( \mathbf{M}^{-1} \) represents the mapping between \( \dot{\theta}(t) \), true angular velocity produced at the handle by the hand and measured directly from the encoder, and \( \tau(t) \), the extra torque required to overcome the inertia of the motor in order to maintain \( \dot{\theta}(t) \). By passing \( \dot{\theta}(t) \) through \( \mathbf{M}^{-1} \) and feeding as additional input to the motor, we make the motor produce the extra torque, which eliminates this inertia as experienced by the hand. This ensures that the torque experienced by the limb while in motion is equal to the phenomenal component of \( \tau_{\text{cm}}(t) \) which is a design variable intended by the experimenter. \( k_c \), a fraction close to but less than 1, is there to ensure stability.](image)
for the inertia of the motor, it will mean that the hand will need to generate some portion of the extra torque in order to overcome inertia. This is still tolerable within practical limits, and furthermore, it can be moderated through other means which we will discuss later. If there is overcompensation, then the handle will in effect begin to drive the hand. As a worse case scenario, the behavior of the handle may become unstable. The compensation gain will typically lie in the upper half of the range $0 < k_c < 1$. Once $M^{-1}$ is implemented, $k_c$ is set to the highest value that does not result in instability or the handle driving the hand.

2.4. Establishing $M^{-1}$: Box–Jenkins model

System identification techniques such as Box–Jenkins are valuable tools for establishing mathematical models of dynamical systems such as our motor-handle apparatus (Box & Jenkins, 1976; Ljung, 1999). There are two key considerations that we must keep in mind in order to get the best out of the Box–Jenkins method. First of all, the Box–Jenkins method uses the input–output data of the system it models. In other words, what it “knows” of the system, is what is contained in the input–output data. Second, for a given order of parameters, the Box–Jenkins method gives the best average linear fit of the input–output data. That is, whatever error that remains between the actual output and the model output is not related linearly to the input signal.

Clearly, the nature of the input signal used during modelling deserves attention. Stochastic signals are smoothly changing random combinations of sine waves and make the most suitable input signals. The bandwidth of a stochastic signal denotes the range of frequencies of the constituent sine waves. For example, a 3.5 Hz-bandwidth stochastic signal consists of sine waves with frequencies ranging from 0 to 3.5 Hz. How do we best choose the bandwidth of the stochastic signal to be used as input? Based on the first consideration above, we would want the signal to have a broad bandwidth, such that the modelling process is properly “aware” of the system. The second consideration tells us that our model will be a linear representation of the system, averaged across the range of frequencies spanned by the input signal. Thus, we would not want to include frequencies that are not of practical interest. The optimal strategy, therefore, would be to choose the bandwidth of the input signal such that it matches the range of frequencies to which the system will be subjected during its normal operation. In our case, this means that the stochastic input signal should consist of those frequencies which are likely to be generated during the interlimb coordination tasks.

We applied a 3.5-Hz bandwidth stochastic signal with a duration of 50 s as torque input to the motor, placed vertically such that the handle motion is not influenced by gravity, and measured the resultant angular velocity. We used the Box–Jenkins method to establish a mathematical model of the motor as the best linear fit between the input torque in Nm and the resulting angular velocity at the handle.

3. Results

3.1. Box–Jenkins model of the motor

The diagram below (Fig. 4) illustrates the high level of accuracy with which the Box–Jenkins model captures the low-pass filtering effect upon the wrist by the motor. The fit between the actual output of the motor and that of the model is at 91.1%. The calculated (discrete-time) model in the $z$-plane, $M(z)$, and its first order continuous (zero order hold) equivalent in the $s$-plane, $M(s)$ are as follows:

$$M(z) = \frac{0.76z - 0.3811}{z - 0.9944} \quad M(s) = \frac{0.76(s + 100.6)}{s + 1.13}$$

As quantified by $M(s)$, the motor does behave like a low-pass filter, with an asymptotic $-20$ dB/decade attenuation from about 1.13 rad/s (0.18 Hz) onwards. As a rule, the closer the positive denominator constant of (a first order) $M(s)$ is to zero, the more pronounced the attenuation effect. Reducing the attenuation effect means compensating the motor to increase this constant (see Nise (2000) for details on discrete-time and continuous-time transfer functions).
3.2. The compensated motor

$M^{-1}(s)$ is equal to the reciprocal of $M(s)$:

$$M^{-1}(s) = \frac{s + 1.13}{0.76(s + 100.6)}$$

As the order of the numerator of $M^{-1}(s)$ is not higher than the order of its denominator, and the root of its denominator is negative, we know that $M^{-1}$ is both realizable as a filter, and is stable, i.e., for a bounded input, it produces a bounded output. Thus ascertaining its suitability, we used it to compensate for the motor as depicted in Fig. 3. We obtained optimal performance (that is, maximum compensation of motor inertia while ensuring stability for all experimental conditions) when the compensation gain, $k_c$, was set to 0.7. Once compensated, the previously experienced sluggishness of the motor vanished. One way to quantify the improvement brought about by compensation is to calculate the model of the compensated motor using the same stochastic input and applying the Box–Jenkins method as before. The discrete-time model, and the continuous-time (zero order hold) equivalent of the compensated motor are, respectively:

$$M_{\text{compensated}}(z) = \frac{0.484z + 0.394}{z^2 - 1.494z + 0.539} \quad M_{\text{compensated}}(s) = \frac{4.7340 \times 10^4}{(s + 24.69)(s + 98.78)}$$

On the basis of the Box–Jenkins method, $M_{\text{compensated}}(z)$ is best depicted as a second order system, i.e., with two denominator constants. Using higher orders does not increase the motor output variance accounted by the model. Given the ratio of the two constants, it is the smaller constant that predominantly determines the attenuation effect of the motor (see Nise (2000) for a detailed discussion on second order continuous-time systems). The smaller constant of the compensated motor being 24.69 and about 21 times larger than the constant of the motor, explains the subjective experience that the sluggishness vanished following compensation. The Bode plots in Fig. 5 (adjusted for unity DC gain) of the motor before and after compensation depict the relative low-pass characteristics. The $-3$ dB point, which is a frequency measure of when attenuation starts, goes from 0.17 to 3.70 Hz. Notice that the $-3$ dB point increases by a factor of 3.70/0.17 equalling 20.5, consistent with our expectation of the basis of the denominator constants of the transfer functions before and after
compensation. Given the 3.7 Hz is at the upper limit of the movement frequencies which the participants are required to produce – 0 to 3.7 Hz range of motion being sufficiently wide to dissociate relative stability of coordination patterns – the motor no longer operates as a low-pass filter on joint dynamics as far as our interlimb coordination tasks are concerned. While the rates of reciprocal movement that can be achieved by segments of low inertia, such as the forearm and hand, may be in excess of 6 Hz (Keele & Ivry, 1987), we are not aware of any instances in which coordinated movements, sustained at frequencies approaching 3.7 Hz, have been reported (cf. Riek & Carson, 2001).

Modelling the motor with and without compensation does quantify the improvement brought about by compensation, and demonstrates the suitability of the compensated motor for the task of generating torque at the handle without attenuating dynamics at the same time – albeit in the relatively abstract language of control theory. We can obtain a more direct measure of how things improve with compensation, and how much or little work the limb has to do to overcome whatever inertia remains after compensation, by means of the reactive torque sensor mounted at the base of the handle. Before we do this, however, we will discuss one final addition to the control strategy, also made possible by the availability of the reactive torque measure.

3.3. Torque error feedback

We mentioned earlier when we were discussing the undercompensation resulting from $k_c$, that we would introduce another way to address it. In principle, the reactive torque sensor enables us to correct for $k_c$, as well as for the model error – the small difference between the actual behaviour of the motor, and the model $M$. It is important to remember that the torque sensor is connected to the motor shaft on one side and to the base of the handle on the other, and measures the differential torque between the two.

Let us go back to our horse and motorized carriage analogy, where the torque sensor would be the rope that binds the horse to the carriage. If the motorized carriage follows the path of the horse perfectly, the horse will not have to pull the carriage and there will be no tension in the rope. If the carriage does not follow the horse perfectly, then there will at times be tension in the rope (the rope only

Fig. 5. Bode plots for the motor before and after compensation. The plots are adjusted for unity DC gain so as to reflect the relative low-pass filtering characteristics of the motor in the two conditions. The –3 dB point with and without compensation are 0.17 and 3.70 Hz, respectively.
captures tension when the horse is outrunning the carriage and not the other way round, whereas the torque sensors measures it in both directions). Now, as the hand moves up and down in the handle generating a particular $\dot{\theta}(t)$, if $M$ is an exact model (and $k_c = 1$) then $M^{-1}$ will produce just the right additional torque $\tau(t)$ for the motor to follow $\dot{\theta}(t)$. With motor matching the hand, the torque sensor will register only the $\tau_m(t)$ component. If, however, $M^{-1}$ does not produce exactly the right additional torque because $M$ is not an exact model (or $k_c \neq 1$), then the motor will not match $\dot{\theta}(t)$. The difference between what the motor should produce to match $\dot{\theta}(t)$ and what it does produce on account of model error will creep into the sensor reading. If we call this torque reading arising from model error, $\tau_e(t)$, then the reading of the sensor will be $\tau_m(t) + \tau_e(t)$.

The reactive torque sensor enables us to correct for the model error because it captures the torque that arises from model error. We obtain $\tau_e(t)$ from the reading of the reactive torque sensor, $\tau_{reactive}$ and subtracting $\tau_m(t)$ from it. We feed it as additional input to the motor as depicted in Fig. 6 in order to drive it to zero.

3.4. Inertial torque experienced by the limb before and after motor compensation

The torque signal $\tau_e(t)$ (i.e., $\tau_{reactive}(t) - \tau_m(t)$) obtained from the reactive torque sensor arises out of the discrepancy between what $M^{-1}$ should produce to overcome the inertia of the motor and what it does produce. In other words, $\tau_{reactive} - \tau_m$ gives us a measure of the remaining inertial effect of the motor as experienced by the hand as it moves. What we could do is get a participant to generate a particular angular velocity $\dot{\theta}(t)$ under typical experimental conditions with and without compensation, and use $\tau_{reactive}(t) - \tau_m(t)$ as a direct measure of the inertial torque as experienced by the hand. The difference between the compensated and uncompensated conditions will give us a direct measure of the improvement brought about by compensation.

The problem with this is that it is practically impossible for the participant to produce comparable let alone the same $\dot{\theta}(t)$ with and without compensation. An experimental requirement which would span almost all typical uses of the apparatus would be the participant generating a sinusoidal displacement profile gradually increasing from 1 to 3.8 Hz over a course of 50 s. One would be hard pressed to find a participant who could meet this requirement when there is no motor compensation. One way to get around this problem is to get the participant to generate $\dot{\theta}(t)$ with motor compensation

\begin{figure}[h]
\centering
\includegraphics[width=0.5\textwidth]{fig6.png}
\caption{The figure illustrates the incorporation of torque error feedback into motor compensation, in order to eliminate the effect of model error and the earlier introduction of $k_c$. $\tau_e$ is equal to measured reactive torque at the base of the handle minus $\tau_m$. $k_c$ is set to 0.9 to ensure stability.}
\end{figure}
turned on, and record the torque, $k_c \tau(t)$, produced by $M^{-1}$ to overcome motor inertia. The reasoning is that in the absence of compensation, the hand would have to produce this torque to generate $\dot{\theta}(t)$. Hence, while $\tau_{\text{reactive}}(t) - \tau_m(t)$ is the inertial torque experienced by the hand in the compensated condition, adding $k_c \tau(t)$ to $\tau_{\text{reactive}}(t) - \tau_m(t)$ will give us the torque that would be experienced in the uncompensated condition. The following figure (Fig. 7) depicts the nearly complete removal of the inertial effect of the motor. Even when the hand is rotating up and down at 3.8 Hz, the remnant peak-to-peak inertial torque it experiences is around 0.14 Nm, a figure that is negligible both with respect to what it (25 Nm, peak-to-peak) would have experienced without compensation, and to the weight of the hand itself.

Fig. 7. This figure shows the angular velocity measured during a trial reflecting typical experimental conditions for which the apparatus is used, where the participant was required to produce a sinusoidal displacement profile gradually increasing from 1 to 3.8 Hz over a course of 50 s, as paced by a metronome. The second plot shows the remnant inertial torque, $\tau_{\text{reactive}}(t) - \tau_m(t)$, experienced by the hand after motor compensation. $\tau_{\text{reactive}}(t)$ is measured by the reactive torque sensor placed between the base of the handle and the motor shaft. $\tau_m(t)$ is the component produced by the motor to compensate for the weight of the handle and impose the torque field intended by the experimenter. The third plot shows the torque component, $k_c \tau(t)$, produced by the motor to compensate for its inertia.
4. Discussion

In this study, we presented an apparatus which enables the experimental manipulation of torque fields around individual limbs, without interfering with the inherent dynamics of those limbs. It is relatively simple to implement and can be constructed with equipment which is readily available and inexpensive. What lies at its heart is the exploitation of the torque generating capacity of an AC servo-motor, while eliminating the motor’s inhibiting effect on joint dynamics.

Ordinarily, the experimenter has limited scope in manipulating torque fields, and the manipulations that are possible result in the introduction of confounds. Given the constancy of terrestrial gravity, one would have to change posture or plane of motion in order to change torque fields acting upon joints, yet these are in themselves important factors that influence the observed patterns of coordination. For example, factors that determine the stability of coordination patterns seem to change completely as we go from the transverse to the sagittal plane of motion (e.g., Baldissera, Cavallari, & Civaschi, 1982; Riek, Carson, & Byblow, 1992; Salesse, Oullier, & Temprado, 2005). Postural changes, on the other hand, alter the inherent characteristics of joints (such as passive dynamics) which then confound the effect of manipulating torque fields about those joints. In a more direct sense, postural changes have also been shown to affect musculoskeletal parameters such as muscle length and moment arms, the motor commands required to generate movement, and the stability of coordination (Carson, Smethurst, Oytam, & de Rugy, 2007). The apparatus described in this manuscript frees the experimenter from these limitations, thus enabling much greater scope in separating empirically the factors that influence coordination.

Concerning perception–action coupling, there is the issue of perceptual and cognitive factors associated with patterns of coordination in distinction to the muscular activity which brings them about (Mechsner, 2004). It is a legitimate question to ask whether a particular coordination pattern that proves to be more stable than others, is so because its perceived directional qualities make it easier to plan by the CNS, or whether its execution is less demanding on the muscular system. Once again, as in our study on moving with a beat (Carson et al., 2009), an apparatus of this sort would be ideal in answering this question experimentally – it makes it possible to manipulate the torque acting upon the limb and hence the muscular load of motion, independently of the direction of motion.

What our discussion has so far focused on is that the apparatus affords an independent means of torque control such that the experimenter does not need to manipulate either posture or plane of motion to alter torque fields about a given joint, and thus avoids the confounding effects of changes in inherent joint characteristics. The experimenter may, nevertheless, be interested in altering posture or plane of motion for other reasons. Changes in joint dynamics resulting from variations in posture or direction may need to be compensated as an experimental control measure. For example, the experimenter may want to study and contrast the effect of gravity on motion under different postural conditions. In this case, it will be important to know to what extent any observed behavioural differences between postural conditions are due to the particular state of joint characteristics at those conditions. The added benefit of the apparatus is that on account of its inertia being much smaller than that of many joints (e.g., wrist, ankle, elbow, knee, shoulder), it can be used to model and compensate joint characteristics in order to minimize differences across different conditions.

Technically speaking, the technique for modelling joint dynamics is in essence the same as that used for modelling the motor itself. We start with the compensated motor and knowledge of the weight of the limb (e.g., hand). The hand is placed in the handle during the modelling process, and the weight of the hand is neutralized by the motor, such that under complete relaxation the hand rests at the midpoint of the experimental range of motion ($\theta = 90^\circ$ from the vertical as shown in Fig. 1). As before, amplified audio feedback of activity of FCR and ECR is presented to participants to aid complete relaxation. A version of the stochastic input signal used to model the motor, appropriately scaled to span the experimental range of motion drives the relaxed hand, while angular position and velocity are recorded. These data are then used to calculate the Box–Jenkins model of the wrist. The inverse of this model can be used in the same way that the motor model is used (see Fig. 3) to generate an additional component to the motor input to compensate for joint dynamics. Should the non-linearity of joint dynamics with respect to angular position or frequency of motion prove to be large enough to
be a concern empirically, then a piece-wise linear modelling approach can be adopted. In effect, rather than one model for the joint, a series of models specific to particular ranges of angular position or frequency of motion can be smoothly concatenated to represent the joint.

Finally, the reader may require us to comment on the possible implications of the apparatus, or of the insights gained from its design on previously published discoveries. When considering the findings of studies on coordination behaviour that have utilized manipulanda to measure limb motion, it may be important to question whether the attenuating effect of the manipulanda on joint dynamics is negligible or not. Likewise, in studies that have used a torque motor without compensation along with manipulanda to alter the dynamics of a joint (e.g., damping or viscosity), it would be important to imagine the implications of the additional and unintended changes on the joint dynamics due to the low-pass filtering effect of the motor-manipulandum system.

On a more positive note, interesting microgravity/hypergravity experiments utilizing parabolic (aeroplane) flights have been reported (Crevecoeur, Thonnard, & Lefevre, 2009) involving single-joint per limb pointing motion for which the use of our apparatus would constitute a direct alternative. Compared to parabolic flights, the apparatus presented here would be cheaper, more manageable (parabolic flights provide short periods of stable gravity, e.g., 20 s, within which individual trials would have to be completed), and more flexible, offering the experimenter a continuum of gravity conditions rather than just two distinct points – zero gravity and 1.8 times terrestrial gravity. A pertinent finding reported by Crevecoeur et al. with respect to this final point is that participants showed distinct types of adaptation for the hypergravity condition depending on whether they were performing the gravity assisted phase of the pointing movement or the gravity opposed phase. As our apparatus is capable of imposing reverse gravity (in this case, the value of interest would be – 1.8 times terrestrial gravity), it would provide the added opportunity of testing whether this reported finding is direction dependent or whether it truly does depend on gravity assistance/impedance. On the other hand, parabolic flights would subject the whole body to the new gravity conditions instead of just the joint(s) in question.

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References


