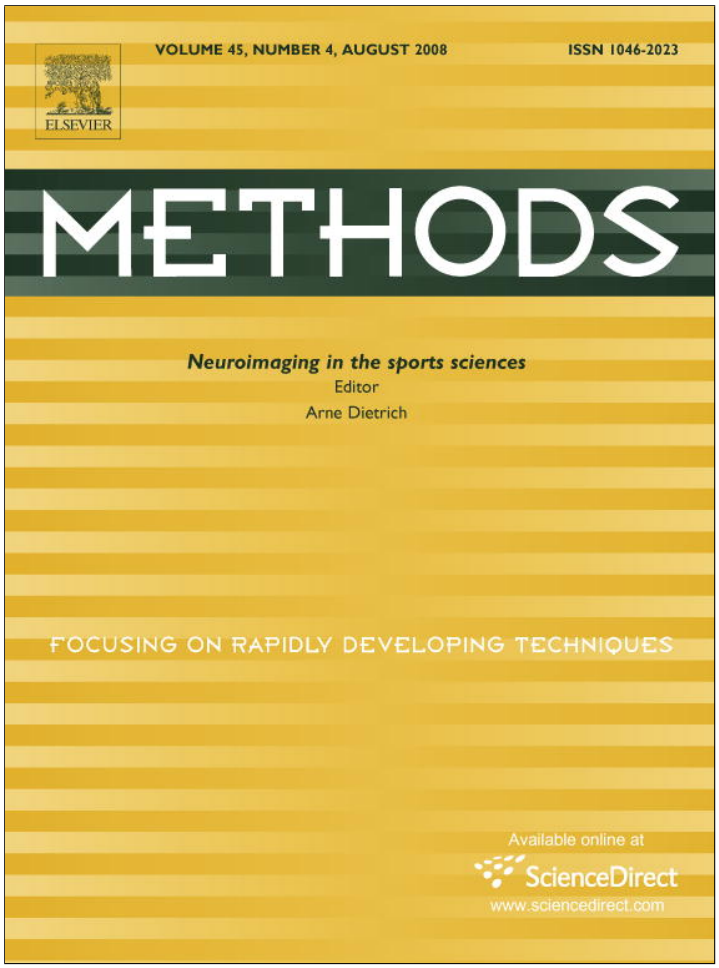

Methods 45 (2008) 279–288

Contents lists available at ScienceDirect

Methods

journal homepage: www.elsevier.com/locate/ymeth

EEG applications for sport and performance

Trevor Thompson *, Tony Steffert, Tomas Ros, Joseph Leach, John Gruzelier

Department of Psychology, Goldsmiths College, University of London, New Cross, London SE14 6NW, UK

ARTICLE INFO

Article history:
Accepted 13 July 2008
Available online 3 August 2008

Keywords:
EEG
Sport
Artifact
Motion
Movement
EEG-biofeedback
Neurofeedback

ABSTRACT

One approach to understanding processes that underlie skilled performing has been to study electrical brain activity using electroencephalography (EEG). A notorious problem with EEG is that genuine cerebral data is often contaminated by artifacts of non-cerebral origin. Unfortunately, such artifacts tend to be exacerbated when the subject is in motion, meaning that obtaining reliable data during exercise is inherently problematic. These problems may explain the limited number of studies using EEG as a methodological tool in the sports sciences. This paper discusses how empirical studies have generally tackled the problem of movement artifact by adopting alternative paradigms which avoid recording during actual physical exertion. Moreover, the specific challenges that motion presents to obtaining reliable EEG data are discussed along with practical and computational techniques to confront these challenges. Finally, as EEG recording in sports is often underpinned by a desire to optimise performance, a brief review of EEG-biofeedback and peak performance studies is also presented. A knowledge of practical aspects of EEG recording along with the advent of new technology and increasingly sophisticated processing models offer a promising approach to minimising, if perhaps not entirely circumventing, the problem of obtaining reliable EEG data during motion.

1. Introduction

Ways in which we can improve sporting performance are of great contemporary interest. A survey in 1997 showed that nearly $12,000 million was spent on ergogenic aids and dietary supplements in the US, with approximately 50% of the general population having reported their use [1]. An alternative approach to understanding performance enhancement has been to study cerebral activity through electroencephalography (EEG). If skilled performing in a particular field is associated with a distinctive EEG profile, this could help us to understand important cortical processes underlying peak performance. In addition, the potential to optimise performance is offered by training an individual's EEG to increase or decrease in a desired direction by self-regulatory techniques such as EEG-biofeedback (also called neurofeedback). A recent article in Men's Vogue claimed that tennis champion Mary Pierce, Olympic gold-medal skier Hermann Maier, and several players on the 2006 World Cup winning Italian football team are all reported to have used neurofeedback to improve their performance [2], although few validation studies have been conducted. In order to establish whether performance enhancement through EEG-biofeedback extends beyond anecdotal reports, well-controlled scientific studies are clearly required. However, our ability to make accurate inferences regarding links between EEG and performance is dependent upon our ability to obtain reliable EEG data. Unfortunately, recording EEG while the subject is in motion is inherently problematic. All of the usual causes of EEG artifact (muscle potentials, sweating, electrode movement etc.) tend to be exacerbated during motion. Furthermore, additional problems such as equipment portability and restriction of the individuals' natural movement are also introduced.

The overall purpose of this article is to discuss the issues of motion-related EEG artifacts within the context of the sports sciences. Firstly, we provide a brief overview of the basic principles of EEG. Secondly, we discuss how previous studies have attempted to apply EEG methodology to sports research. Thirdly, the problems that EEG recording in sports typically presents will be discussed along with practical, technological and computational methods for tackling these problems. Finally, we discuss whether attempts to alter an individual's EEG through neurofeedback have successfully resulted in performance enhancement.

2. Basic principles of EEG

The electrical activity of neurons in the brain produces currents that reach the surface of the scalp. EEG provides a non-invasive method of recording the voltage differences of these scalp potentials. These potentials are created by both cerebral sources and unwanted non-cerebral artifacts which tend to be exaggerated during movement. The EEG signal is transmitted from the scalp electrodes to a differential amplifier in order to amplify the microscopic

* Corresponding author. Fax: +44 207 919 7873.
E-mail address: t.thompson@gold.ac.uk (T. Thompson).

1046-2023/$ - see front matter © 2008 Elsevier Inc. All rights reserved.
doi:10.1016/j.ymeth.2008.07.006

potentials severely attenuated by their passage through the skull. This signal is continuously sampled at a high rate (typically 256 Hz but often more) to provide a high temporal resolution. An analogue band-pass filter is used to filter the raw EEG signal and typically possesses a lower cut-off of 0.5 Hz and a higher cut-off of 50 Hz. The 50 Hz filter helps eliminate electrical noise originating from 50/60 Hz mains power. These filters also affect the processing of nearby frequencies so care must be taken to ensure the cut-off frequencies do not lie too close to the frequencies under investigation. The default cut-offs pose no problems in the sports sciences as the low to mid range frequencies (e.g. 4–20 Hz) are normally those of interest.

After amplification and filtering, the EEG signal is (in modern digital systems) relayed to a computer where it can be processed as continuous data and, if desired, its spectral parameters compared with some criterion measure. This is the approach adopted by EEG-biofeedback training in sports and other performance domains which rewards desirable changes in specific frequency bands. An alternative approach is the study of event-related potentials (ERPs). These usually consist of data epochs of short duration reflecting the cortical response to an external stimulus. In order to offset data noise, many ERPs (often hundreds) are averaged to provide a favourable signal-to-noise ratio.

The typically wave like appearance of the EEG signal reflects the rhythmic activity of underlying synaptic processes. This rhythmicity is thought to reflect the synchronised activity of large neuronal assemblies possibly driven by thalamic pacemaker cells [3]; although the simplicity of this interpretation has been questioned [4]. Anatomically distinct cortical areas produce a variety of different rhythms which are observed as a composite EEG signal. Fourier spectral analysis is typically used to decompose this signal into its constituent frequency bands and to compute the amplitude of each band. These bands have been historically categorised as delta (<4 Hz), theta (4–8 Hz), alpha (8–12 Hz) and beta (13–30 Hz), although alternative classifications have also been employed [5]. Slower waves such as delta are typically associated with sleep while faster beta waves are associated with wakefulness and mental activity. Alpha has been linked to a 'relaxed focus or mental readiness'. An increase in alpha activity is often the goal of EEG-biofeedback training aiming to improve sporting performance [6] through increasing the user's ability to remain focused thereby filtering out distracting stimuli, thoughts or emotions. In addition to spectral analysis, more complex analytical techniques have been developed including source localisation methods such as low resolution electromagnetic tomography, or LORETA [7], which aim to identify the original sources of cortical oscillations. Such methods are described later in this paper.

2.1. Recording methods

EEG measurement entails the attachment of electrodes to standardised locations on the scalp. These electrodes are generally made of highly conductive silver or silver chloride (Ag/AgCl) although other metals such as tin, gold and platinum are also used. Non-metallic material such as carbon fibre can also be employed to allow compatibility with other neuroimaging devices such as MRI. Electrodes are attached to the skin using conductive paste with impedances generally kept below 5 kΩ. Prior to attaching the electrodes the skin is usually prepared with an abrasive paste such as Nu-Prep to reduce skin impedance [8,9]. The number of active electrodes can range from one, which is sufficient for neurofeedback training, to multiple electrodes necessary for source localisation with the number of electrodes typically varying from 20 to 128 [4]. Electrode placement is standardised to aid interpretability from one laboratory to another. The standard method of electrode placement is the international "10–20 system" [3] depicted in Fig. 1.

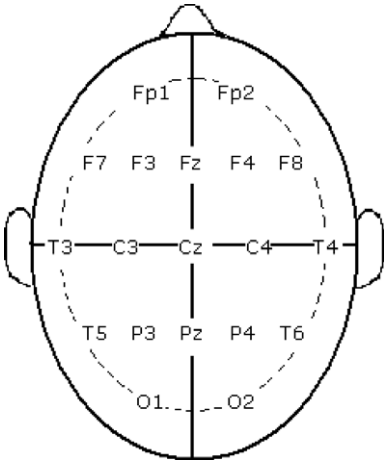

Fig. 1. "10–20" system of electrode placement. F = frontal, T = temporal, C = central, O = occipital, P = parietal. Odd numbers = left hemisphere, even numbers = right hemisphere.

A differential amplifier measures the voltage difference between inputs from the active and reference electrodes, with the resulting signal amplified and displayed as a channel of EEG activity. A signal that is common to both inputs is thus automatically rejected in what is known as common mode rejection (CMR). 'Noise' shared across electrodes is thus effectively eliminated leaving only the (hopefully neural) activity specific to the active electrode. In sporting applications, the reference tends to be from electrodes placed on the mastoid (the bone behind the ear), occasionally the ear lobes or the average of all (common average montage) or surrounding (Laplacian montage) electrodes in multi-channel setups.

3. Applications of EEG methods in sports research

Given the difficulties of recording EEG during movement, researchers have perhaps unsurprisingly explored alternative ways to apply EEG methodologies in the sports sciences. One approach is to record EEG outside of the execution of the sporting task itself, in order to assess long term clinical outcomes associated with injuries sustained by athletes or to understand pre-task cortical processes. The use of 'imagined' sporting actions as a convenient proxy for real sporting activity is an additional favoured approach. In some cases it is also possible to record EEG during the actual critical movement phase, as in simulated diving conditions or the use of stationary cycling equipment where electrical interference is kept to an acceptable minimum. The following will provide a brief review of studies employing these approaches.

3.1. Pre/post measurement of EEG

The first reported use of EEG methods in sports science surfaced in the early 1950's with investigations into boxing [10,11]. Researchers in this field have introduced two distinct themes. Firstly that of studying pre-post combat differences in electro–cortical activity, and secondly examining the differences between healthy adults and professional boxers subjected to repeated head injury over the long-term. Due to the evidence linking EEG findings with symptoms of brain damage, the case has been made for changes to boxing rules, conditions, and health screening [12]. A related line of inquiry has assessed long term outcomes for footballers following repeated minor head trauma caused by contact between the head and the ball [13,14]. The authors of these studies

explored associations between self-reported symptoms such as headaches, dizziness, irritability, memory impairment and neck pain, and abnormal patterns of EEG activity. The findings have come under some criticism on methodological grounds [15], and as such provide only suggestive evidence that EEG methods can be used to signify brain injury in football.

With respect to the study of pre-task cognitive activity, the process of taking aim in target shooting sports presents ideal conditions for EEG recording, as it features a period of motionlessness whilst the target is being attended to prior to firing. This pause in movement has provided researchers with sufficient scope to describe optimal patterns of cortical activity for taking aim in several sports: archery [16,17], golf [18,19] and rifle shooting [20–22]. Applications in this setting have been able to define predictors of optimal performance in two ways, in differences between expert and non-expert performance, and in pre-shot EEG differences between successful and unsuccessful shots. This is a salient point as it provides the basis for EEG analysis (real time or post hoc) to be incorporated into athletic training, as means of cognitive augmentation, indicating relationship between the athlete's current state of cortical activation and the established criterion for optimal performance.

3.2. Imagined movement

Another way in which the inaccessibility of EEG during movement measures has been addressed by researchers is to trigger neural activity in the motor cortex by means of imagined movement. This is considered perhaps the least ecologically valid means by which to work within the constraints of EEG recording in the context of physical action as, of the methods described so far, it is the most far removed from the process itself. Nonetheless, motor imagery such as imagined 100 m swimming [23] or imagined training competition [24] discloses some differential effects on alpha band EEG signatures as left occipital and pre-central areas or overall mean alpha frequency calculations, respectively. In more recent times Brain Computer Interface applications have interpreted imagined walking and stopping, as recorded from the motor cortex [25], as controllers for navigation around virtual environments. At this stage however, the inferences drawn from the EEG signal develop very rudimentary movement switches that barely begin to approach the complexity and diversity with which actions typically encountered in sports are observed [26].

3.3. Simulated sporting environments

EEG methods have been used to study impairments to cortical function associated with the environmental conditions in which some sporting activities take place. In diving, High Pressure Nervous Syndrome is characterised by symptoms such as intention tremor, ataxia, motor weakness, sensory symptoms, vertigo, nausea and reduced memory [27]. EEG studies have explored the neural correlates of these symptoms by obtaining reliable measurements through the use of simulated high pressure environments, making it possible to obtain accurate measurements in the conditions being examined [27–29]. The effects of high altitude on mountaineers have also been investigated by means of EEG methods when exploring associations with the symptoms of Acute Mountain Sickness (AMS) such as dizziness, headache, confusion and cerebral edema. Resting EEG measurements were made at base camp and high altitude levels, and have established EEG predictors of AMS as symptoms develop from sub-clinical into clinical [30]. EEG has also been used to assess the sleep problems experienced at high altitudes, diagnosing reduction in stage 4 sleep in comparison with sea level measures [31]. Medication for assisting sleep at high altitude has been validated using EEG recordings to increase sleep quality by increasing slow wave sleep and stage 4 sleep in comparison with placebo controls [32].

3.4. EEG recordings of in-task cortical activity

The use of cycle ergometers has afforded a favourable enough signal to noise ratio to enable measures of electro–cortical responses to physical exercise whilst it is taking place. This affordance stems from the minimal head movement resulting from the stability of the equipment. Although some studies have indicated reduced cortical arousal during exercise, others have demonstrated increased arousal. Short-term exercise of moderate intensity has, for example, resulted in reduced activation in the prefrontal cortex [33] and decreased cognitive performance [34]. In other studies, however, exercise has resulted in increased cortical activation, with increases observed in the P300 amplitudes of ERPs suggestive of a facilitation of cognitive processing [35–37]. The apparent discrepancies across the findings of such studies may be attributable to variation in methodological factors such as the intensity and duration of the exercise as well as the physical fitness of the participants [38].

4. Artifact reduction: practical and technical approaches

A substantial problem in EEG is obtaining 'clean' data on cerebral activity, i.e. uncontaminated by non-cerebral artifacts. Physiologic artifacts (e.g. muscular activity and eye blinks) are generated from the body, while extraphysiologic artifacts (e.g. environmental electrical noise) originate from sources outside the body [39]. Physiologic artifacts tend to be a particular problem when recording EEG from a subject who is in motion. This may account for why studies of EEG in sports have generally been confined to disciplines involving relatively minimal head movement such as golf, stationary bike cycling and rifle shooting. Nevertheless, two complementary approaches exist that can substantially reduce or eliminate artifacts. The first involves minimising movement artifacts during the recording itself. The second requires subsequent signal processing of the data via computational methods to remove artifacts. The following section will consider some of the artifacts especially pertinent to the sports sciences, how to identify such artifacts, and methods to minimise their occurrence (for the topic of EEG artifacts in general the interested reader is referred to a number of good sources [3,40–42]). Computational methods for artifact removal will be considered in the subsequent section.

4.1. Muscle artifact

Muscular contraction elicits myogenic potentials that can represent a major source of EEG artifact. Sports that involve frequent and intense muscular contraction thus tend to elicit a high degree of electromyographic (EMG) artifact. EMG can exhibit an amplitude of around 100–1000 μV, considerably greater than that of EEG (around 10–100 μV) [43]. Consequently, muscular activity can obscure neural potentials altogether. This has historically been a problem with ambulatory monitoring of EEG in epilepsy, where EMG spikes can obscure the detection of epileptic spikes [44]. Similarly, the fact that muscle artifact can completely obscure EEG activity can potentially limit EEG applications in the sporting domain.

What can be done about EMG artifact? Fortunately, it is usually easy to distinguish between substantial EMG and EEG from the raw signal morphology, spectral distribution, and scalp location. EMG consists of a series of spiked discharges from underlying motor units. The frequency of the discharges can range from 20 to 1000 Hz, depending on how many muscle fibres are recruited

and the degree of muscular contraction [43]. However, the dominant energy is in the 50–150 Hz band. In contrast, more than 90% of the EEG's spectral power lies within 1–30 Hz frequency. If the brain activity of interest lies below 15 Hz, simple use of low-pass filtering and/or avoiding the directly contaminated electrodes may facilitate adequate signal detection. Muscle artifact also tends to occur in specific places and these should be examined. Scalp locations most affected are the temporal areas T3 and T4 which lie in close proximity to the temporalis muscle. Artifact here thus tends to reflect jaw movement or tension. This is illustrated in Fig. 2 with irregular activity at T3 with increased power indicated at this location on the topographical scalp map. Jaw tension is a particularly common muscle artifact and, if present, it may be productive to show the subject the effects of muscle movement and tension on the EEG prior to recording and allowing them to learn to reduce their impact. Chewing should also be discouraged.

Additionally, frontal sites Fp1, Fp2, F7 and F8 lie in the region of activity of the frontalis muscle (the 'frowning' muscle) of the forehead. Activity in this region in illustrated in the topographical map of Fig. 3, which also reveals possible neck tension emanating from activity in posterior leads O1 and O2. In addition, the spectral map indicates that peak amplitude at F3 occurs at a high frequency of around 28 Hz.

A common way of eliminating overt EMG (and other types of) artifact is to simply reject the contaminated portions of the EEG. However, when the degree of contamination is considerable, as can be the case when physical exertion is high, rejection can result in a considerable loss of hard-earned data; perhaps leaving too little for meaningful analysis. In this instance, more advanced post-processing methods such as Independent Components Analysis (ICA) can be attempted to separate the EMG signal from the raw EEG signal. ICA has shown some promise in isolating muscle artifact. ICA and other methods are described in the later section on computational methods.

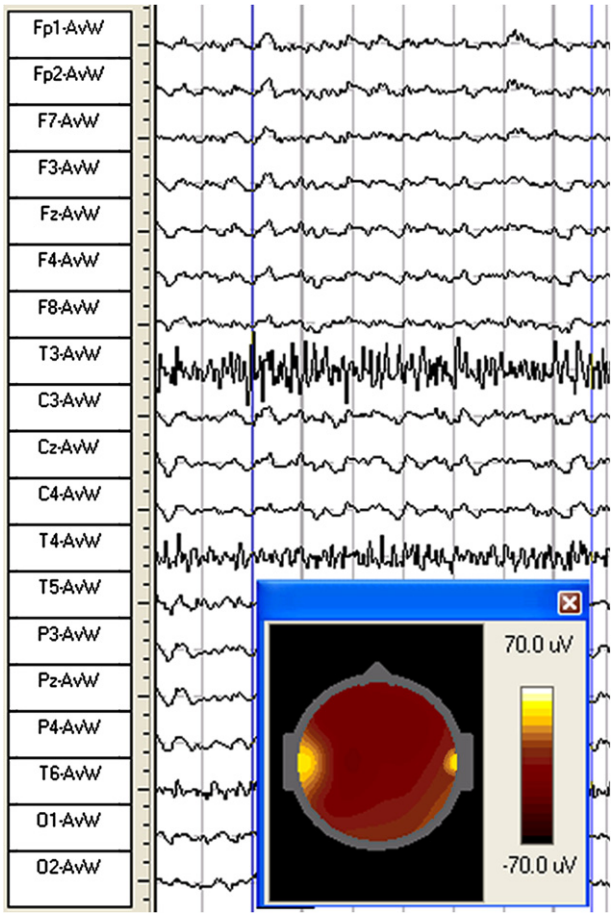

Fig. 2. EEG signal and topographical map indicate jaw tension.

4.2. Skin artifact

Sodium chloride and lactic acid from sweat glands in the scalp can react with the metal of the electrode to alter impedance and thus signal amplitude. If this occurs differentially across active and reference electrodes an impedance mismatch naturally results [45] which can result in large baseline sways. Sports that involve sustained physical exertion are naturally more likely to cause sweating and produce these types of artifacts. This problem can be exaggerated with the use of EEG caps or in bald subjects with no hair to help absorb the sweat. Such artifacts are generally recognised by very low frequencies of below 1 Hz and are thus are often easily distinguishable from the mid-range frequencies usually of interest in sports research. Nevertheless, it is good practice to try to reduce the influence of these artifacts at source. Generally, any steps that are likely to lower body temperature are likely to minimise their appearance. The use of a cool air-conditioned room, where possible, and the avoidance of excessive layers of clothing may help. Where feasible, frequent breaks may also help to keep body temperature low. For similar reasons, if one is recording swimmers outside a pool it is important to make sure their hair is properly dried. All EEG amplifiers should be electrically isolated or wirelessly remote from any mains source of electricity, so not to represent an electrical safety hazard in a wet environment. EEG amplifiers use CMR to reject signals common across inputs. However, impedance mismatches between electrodes can result in a common signal (such as 50 Hz noise) producing *different* voltages across amplifier inputs. This mismatch can result in a distortion of the EEG signal (see Freye [46] for a good account of how the size of the mismatch is related to the magnitude of the signal distortion). It is important therefore to monitor impedances to ensure that differences across active and reference sources are minimal, ideally by ensuring all impedances are kept low.

4.3. Electrode movement

Any movement which disturbs the contact of the electrode with the scalp can result in a sudden increase in electrode impedance [3] resulting in a dramatic change in the EEG signal. Overcoming this issue may be one of the biggest challenges; especially in sports involving a high degree of motion or where the sporting action of research interest necessitates gross motor movement. While electrode movement is easily detected on the EEG signal, contaminated EEG from frequent movement can produce a great deal of data loss. Great care must be taken to ensure a consistent low impedance contact with the skin. Standard ear clip electrodes should be sufficient for recording EEG in relatively stationary sports like target shooting. Self-adhesive pre-gel disposable electrodes that stick to the mastoid (and in bald subjects to the scalp) can be quickly and easily administered and are less likely to come loose in more active sports (see Freye [46] for a well-illustrated description of electrode types). A more secure method is to glue the electrodes firmly to the scalp with an adhesive conductive gel. Epilepsy/sleep labs have traditionally used collodion, a commonly used one being EC2 Grass-Telefactor paste [47]. It should be pointed out that this method of adhesion is also more cumbersome and disliked by the subjects because of the residue left on the scalp, even after removal of the collodion with acetone [48]. A constant pressure on the electrodes is necessary to prevent them from moving horizontally or vertically. This can be aided with the use of a tight elec-

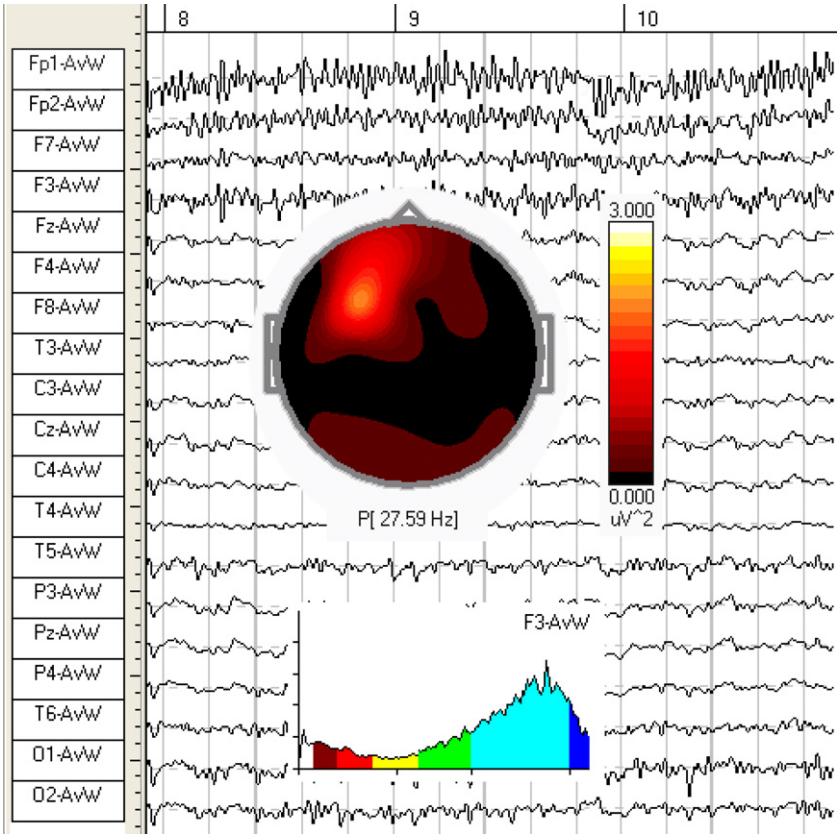

Fig. 3. EEG topographical map and spectral analysis indicate possible frowning and neck tension.

trode cap, or an elastic gauze or head net when the number of electrodes is relatively low. A Lycra cap can be fitted and aligned with minimal effort, with a 19 channel cap taking around 10–15 min to apply. Securing the leads may minimise the gravitational and rotational forces pulling on the electrodes.

Active electrodes where a pre-amp stage is mounted directly to the electrode on the scalp and magnifies the signal before sending it to the main amplifier, can greatly reduce cable movement artifacts and improve the signal quality in all sports activity and are strongly recommended. Amplifiers with high input impedances can reduce the amount of skin preparation needed whilst maintaining the signal quality [45]. Current development of micro-spiked electrodes [49], that can avoid pressure-induced skin potential changes caused by electrode motion or skin stretching, and super high impedances microchip mounted dry electrodes may hold the promise of a more stable EEG signal in the near future.

4.4. Eye movement

Eye movement is a universal source of artifact and can be precipitated by both eye blinks and lateral eye movements. The potential difference between the cornea and retina is larger than that of cortical potentials [50]. During an eye-blink the eyeball turns upwards. This tends to primarily affect the frontal electrodes, with a large positive deflection seen at Fp1 and Fp2 with a peak amplitude of around 50–200 μV lasting 200–400 ms. If the peak is particularly large other electrodes can also be affected. Lateral eye movement is recognisable in the fronto–temporal areas as sharply contoured potentials that are out of phase [3]. Some research has indicated that eye blink rate may not increase during physical exertion [51,52], and therefore this type of artifact may not be any more prevalent during high activity sports. In fact, there is some evidence that increased visual load decreases the rate of eye blinks [53–55]. These types of artifacts may therefore not be any more common in the large number of sports where visual processing demands are high.

As with EMG, eye blink is easily recognised in the EEG as its raw signal morphology and amplitude have a distinctive pattern, contaminating delta (1–4 Hz) and theta (4–8 Hz) bands predominantly at frontal sites (see Fig. 4). Modern blind source separation techniques such as ICA can ameliorate many of the problems caused by eye blinks. It is often wise to record a short test artifact baseline (where the subject attempts to produce various artifacts) to assist in their later identification, by asking the subject to blink their eyes a few times, as well as clenching their jaw and tensing their neck.

4.5. ECG artifact

The electrical activity of the heart is measured by the electrocardiogram (ECG or EKG). The electrical field from each cardiac pulse is very large and can be measured up to a metre away from the body. ECG is more likely to be seen in people with wide necks (such as weightlifters) but it generally does not pose a problem as it tends only to contaminate the low frequencies of around 1–2 Hz. This artifact can be common in channels connected to the ears. Most EEG amplifiers reserve an input for ECG recording, ensuring this artifact is easily recognised. The rhythmic and distinct morphology of ECG also means that it is generally easily removed using the post-processing computational methods discussed in the next section.

4.6. Respiration artifact

Respiration artifact arises from the rhythmic body movement of inhalation and exhalation and may be initially observed as high amplitude deflections with a delta wave like frequency. This

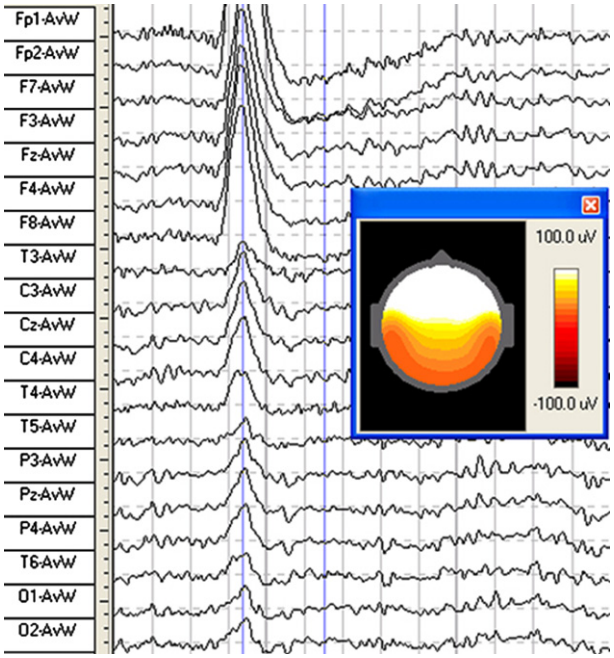

Fig. 4. EEG and topographical map indicate eye blink.

frequency may be expected to increase depending on the aerobic demands of the sport and the fitness of the athlete. As with ECG, one channel can be devoted to respiratory movements which can be measured by a stretch sensitive device worn around the chest or abdomen. These types of artifacts are highly suitable for removal by post-processing methods.

4.7. Tongue movement

Glossokinetic artifacts are created by the potential difference between the tip and base of the tongue and give rise to slow potentials. This type of artifact usually does not occur frequently enough for its removal to cause a significant loss of data. Furthermore, unless systematically co-occurring with the sporting action under study, glossokinetic artifacts may be good candidates for removal by post-hoc computational methods.

4.8. Electrical interference

Electrical noise from the environment is normally eliminated by common mode rejection as previously described. However, if a large discrepancy exists in the impedance (quality of connection) between electrodes, noise will not appear common to both electrodes and will not be successfully excluded. Electrical noise artifact is most notable at the 50 Hz frequency (Europe) or 60 Hz in the US. Ensuring a robust and good quality connection and checking impedance online may help to minimise this artifact. To reduce electrical noise, active shielding can be used, where a signal is passed down the outer shielding layer of the cable to block any external electromagnetic interference. This may be of use in a hostile electrical environment such as within a Formula One racing car.

4.9. Restriction of mobility

Although not an artifact as such, one of the major problems in neuroimaging in active sports is the enforced restriction of motor movement. Fuelled by recent hardware advances, this is the area in which EEG exhibits substantial advantages compared to other neuroimaging technologies. EEG equipment is comparatively cheap, portable and light, and offers the real possibility of measuring neural activity in a real live sporting environment outside of the laboratory.

Portable amplifiers can record multi-channel EEG directly onto a removable flash card for up to 48 h [56] which can later be transferred to the PC for analysis, eliminating the need for cables. These amplifiers can simultaneously record EMG, ECG, EOC, HR, RSP (respiration) and GSR (Galvanic skin response). Developments in wireless technology have also seen the introduction of battery-powered amplifiers capable of transmitting EEG information to a PC or pocket PC in real time [57], allowing for the possibility of EEG-biofeedback during sports training. A 4-channel wireless amplifier can weigh less than 200 g and can be strapped to the small of the back with the cables secured to the head with a sweat band. This can allow the subject relatively unrestricted movement within a radius of around 30 m. With simultaneous recording of sound it is possible to time-lock key neural event events to sporting actions such as a trigger pull or ball contact. As described earlier, remotely transmitting electrodes with active chips are currently under development, and hold great promise for the future in entirely eliminating the need for cables. Battery free systems powered by body heat and ambient light are also currently under development [58].

5. Artifact removal: computational methods

The previous section described ways to minimise movement artifacts present in the recording itself. A second, complementary approach involves separating neural signals and artifacts by post-processing of the data through computational methods.

The choice of post hoc data analysis initially depends on the signal dimensions (i.e. the number of channels), which directly impacts on the relative strength and weakness of the method. Multi-channel data carries more information; hence it is statistically more robust and thus more reliable. However more channels means more data, so blind source separation techniques that, by definition, require multiple channels (such as ICA) involve more extensive experimental setup, as well as a higher computational load; a factor that usually limits them to offline processing. A potential workaround is to have only one or a few electrodes on the region(s) of interest, and concentrate on regions and/or frequencies that are less susceptible to artifact. For example, the central sensorimotor regions are located furthest away from cranial muscles and are also less affected by eye-blink artifacts than frontal regions. Basic digital filtering (e.g. finite impulse response) may also be useful in isolating particular frequencies of interest that are more robust to artifact, such as high theta and alpha (6–13 Hz) which are least affected by both low frequency (movement, blinks and sweat) and high frequency (muscle and electrical noise) artifacts. Another alternative is to apply autoregressive frequency analysis (such as the Yule–Walker algorithm [59]), which is reported to be more stable under movement and noise artifacts than conventional Fourier-based methods, which are prone to spectral leakage and poor performance under conditions of low signal levels (10's of μV) and can result in a low signal-to-noise ratio.

Nevertheless, perhaps the greatest modern advance in artifacting (and signal processing in general) has occurred with the advent of independent component analysis, or ICA, pioneered by Bell and Sejnowski in 1995 [60]. To the best knowledge of the authors, there is still surprisingly little if any published research on the use of ICA to remove large artifacts due to motion during sport or physical exercise. Its most recently reported application in respect has been in correcting task-related movement during fMRI prompted by the extreme sensitivity of this method of neuroimaging to movement

[61]. Nevertheless, ICA does offer great promise as a technique for artifact removal in exercise research. ICA is a higher order statistical method developed to extract individual signals (referred to as components) from mixtures of signals, based on the assumption that different physical processes (referred to as sources) will generate unrelated signals. One methodological caveat is that since the aim of ICA is to separate underlying 'source' signals considered to be 'statistically independent' over time, it requires a relatively large amount of data in both length (EEG samples) as well as channels (the number of sources it yields is directly restricted by the number of recording electrodes used [62]). Even on modern day computers, common ICA algorithms may take from minutes (for individual EEGs) up to hours (for very long EEG or if analysed en masse) to complete. In addition to the assumption of independence of source origins, common ICA algorithms (e.g. InfoMax, FastICA) operate under two further assumptions. Firstly that the underlying sources must exhibit non-Gaussianity, i.e. they must be non-normally distributed (other techniques do not require this assumption (see [63]), although these are not discussed in detail here). Secondly the sources should be stationary, that is to say, they should each have a fixed location throughout the recording. Recent years have seen a remarkable proliferation of ICA related articles with successful applications described in reference to both artifacting [64] and EEG source modelling [65]. Evidence for the former includes artifact removal of muscle [66] contamination, eye blinks and movement [67,68], noise [67,68], as well as cardioballistic phenomena [69]. Fig. 5 illustrates how ICA was used to extract an eye blink component from the raw EEG in data we recently recorded from a sample of contemporary dance performers.

It should be borne in mind that different ICA algorithms (popular ones include Infomax, SOBI, FastICA and JADE) are likely to best detect specific types of artifacts. The JADE algorithm, for example, may be particularly effective for tackling muscle artifact [66]. Complete artifact elimination may therefore require selection of one or more ICA algorithms. The greatest advantage ICA has over conventional artifacting methods is in the fact that an artifact 'component' can simply be linearly subtracted without theoretically incurring any loss to the remaining EEG data occurring simultaneously with it. The spreading popularity of ICA cannot be mentioned without honouring the Matlab toolbox EEGLAB [70], a freely available open source research software (http://sccn.ucsd.edu/eeglab/), and the first to implement ICA (via the Infomax algorithm) for general EEG/ERP analysis and artifacting. A strength of this toolbox is that it also enables the user to identify and cluster matching components (or artifacts) between different subjects based on their scalp map, dipole projection (DIPFIT plugin), spectral power or ERP characteristics. In light of this, and given that ICA typically decomposes spatially fixed and physiologically plausible EEG sources, an effective way to approach artifact-rich data is not to painstakingly remove the artifact cocktail but rather concentrate on extracting the EEG source components themselves. Ironically the motivation for removing artifacts is primarily to get a stable record of the underlying independent EEG, yet this is what ICA is already made to do and does best. Thus, isolating and clustering matching EEG components of interest [71] (frontal midline theta, mu rhythm, or parietal alpha for instance) across subjects not only circumvents complicated artifacting but also minimises the variability and error involved when comparing electrophysiological recordings of different individuals and conditions, as there inevitably exist some individual differences in EEG cap placement as well as physical location/orientation of EEG source dipoles. Notwithstanding, ICA presents an apparent trade-off between the dimensionality of the EEG data (i.e. the number of channels) which when greater, yields more components (thus more complex as well as accurate information), and the practicality of experimental setup or the computational load. Part of the purpose of artifacting is to remove disruptive (or uninformative) data that further adds to the complexity of the EEG. For a fixed number of channels however, some of the components are 'wasted' on the artifacts and hence the remaining cerebral components are not resolved as clearly. An effective compromise is to use an intermediate number of channels (e.g. 21 or 31), perform ICA, subtract the major artifactual components and then run ICA *again* on the remainder of the data. This conveniently restores the maximum number of available components, without the subtracted artifacts.

5.1. Future directions

In the quest for ever-more refined tools to probe the signals of the brain, there is new evidence of work in progress with regards to

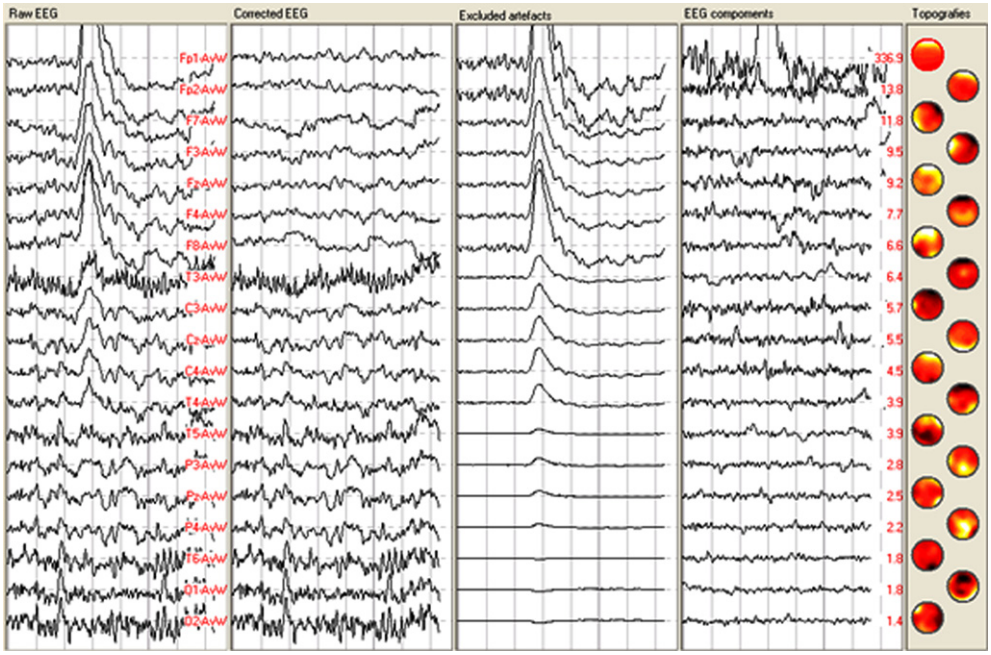

Fig. 5. Removal of eye blink artifact using ICA. Columns show the raw EEG, corrected EEG and excluded artifact, along with EEG components and topographies.

further improving ICA, with a so-called semi-blind approach [72]. By effectively imposing temporal or spatial constraints on the underlying source mixture via such methods as LIANA [73], or by wavelet enhanced thresholding [74] which more faithfully preserves spectral amplitude and coherence. Other research is centred on so-called automated [75,76] artifact removal, which makes use of statistical parameters to automatically classify multiple types of artifacts and remove them without the need for human assistance. Lastly, another promising area which will no doubt prove valuable is *online* real-time artifact removal, which includes both ICA [77] and faster traditional methods [78]. This could be used to extract artifacts a priori during recording itself, making it especially useful for potential EEG neurofeedback applications in sports enhancement (e.g. training alpha desynchronisation while golf-putting [18]).

6. Neurofeedback and performance enhancement

A discussion of EEG methodology in the sports sciences would seem incomplete without some discussion of neurofeedback. Previous research has shown that the EEG of expert sportsmen shows distinct differences relative to non-experts [6,17,79,80]. If a causal link is assumed, neurofeedback offers the potential to provide performance improvements by training an individual's EEG. With conventional neurofeedback, specific components of the EEG spectrum are fed back to the individual in real time using an online feedback loop in the form of either audio or visual information. Visual feedback is often in the form of a moving bar with the amplitude of the selected EEG frequency band represented by the size of the bar, and with the participant aiming to increase or decrease its amplitude as instructed. This may be accompanied by auditory feedback to indicate a point scored. The aim is to train the individual to gain learned control over a particular component of brain activity. Typically a bar representing the chosen frequency band to be enhanced will be increased, while simultaneously higher and lower bands are inhibited and their respective bars reduced in amplitude. The visual feedback may depict a virtual reality performing space to provide ecological validity and transfer learning more effectively to the performance context, as in the case of our current studies of acting performance [81]. Here, as the participant learns to regulate cortical activity the illumination in the theatre auditorium is raised or lowered.

Despite numerous anecdotal reports attesting to its efficacy [2], there are actually very few well-controlled studies that have directly examined the ability of neurofeedback to produce improvement in measures of sporting performance. In fact, to the authors' knowledge only two studies to date have appeared in peer reviewed journals. Landers et al. [16] set out to determine whether neurofeedback could improve performance in 24 skilled archers. One group received peak performance neurofeedback training with reward provided for low frequency activity in the left hemisphere. The rationale for this type of training is based on EEG studies demonstrating more slow frequency activity in skilled marksmen in the left hemisphere (with the most pronounced differences in the left–centro–temporal–parietal areas [82]). In the Landers study, the neurofeedback group showed significant improvements in shooting accuracy. No performance improvements were observed in an 'incorrect' neurofeedback (low frequency activity in the *right* hemisphere rewarded) or a control group, with the former in fact showing a significant deterioration in performance. A more recent study by Arns et al. [83], examined the effect of neurofeedback vs. no neurofeedback on putting performance in golf. The criteria for frequency band reward in the neurofeedback training were based on each participant's individual EEG profile during successful putts prior to neurofeedback training. The overall percentage of successful putts was significantly greater after neurofeedback was administered.

Although there is a paucity of studies directly examining the effects of neurofeedback in sport, a number of controlled studies have examined the impact of such training on performance measures outside of the sporting arena. Neurofeedback has elicited positive changes in the domains of memory, attention, creativity and mood [84–87], which has promising implications for peak performance training in sports. Neurofeedback has a long history of use as a clinical application [88–90], although such applications are not discussed here. Probably the largest body of neurofeedback validation studies has been conducted with the aim of enhancing cognitive performance [91]. The underlying rationale for these studies is that a specific cognitive function can be enhanced by training the frequency most closely associated with this function. Studies of neurofeedback and cognition appear to be primarily devoted to examining the relationship between beta (13–30 Hz) and focused attention, and it is these studies that may have the most relevance to the sporting world. It should be noted that an extensive coverage of such studies is beyond the scope of the current paper, with a more detailed review provided by a number of good sources [6,91,92].

Fast wave training components of sustained attention, such as omission versus commission errors, have found to be differentially influenced by adjacent bands; e.g. 12–14 Hz termed the sensory motor rhythm band (SMR), and 15–20 Hz referred to as the low beta band [86,93]. In one such study, improvements in sustained attention along with verbal working memory were observed only in the SMR group [84]. The SMR band has been of particular interest where, through the reduction in excitability of the sensory motor cortex, performance is characterised by a sustained and relaxed attentional focus, an enlarged working memory space, and a more modulated performance with greater readiness to respond and a more efficient performance overall [94]. In a preliminary study, trainee eye surgeons receiving SMR training performed more efficiently with less time on task, yet with slightly longer pauses between tasks [95].

Fast wave training may be contrasted with slow wave training with eyes closed where auditory feedback is contingent on production of theta (4–8 Hz) and alpha activity (8–12 Hz). The aim is to increase the theta to alpha ratio by making the sound associated with theta particularly conducive to relaxation (e.g. the sound of waves breaking on the shore). About two thirds of individuals begin with alpha higher than theta, and to set the relaxation process in motion alpha may be rewarded with a lower threshold than theta at first. Phasic increase in theta and alpha may also be rewarded with separate sounds such as a resonant gong. Theoretically the aim is to induce relaxation to a state of hypnogogia which has historically been associated with creative insights [96]. Slow wave training has benefited competitive ballroom dance performance in a controlled investigation [97]. Professionally significant improvements were obtained with alpha/theta training in dancers who went on to win the UK university championship. Interestingly another biofeedback procedure—heart rate coherence training—was equally effective in improving performance overall, with particular impact on technique, whereas slow wave training impacted on timing. Slow wave training has widespread implications. It enhances feelings of confidence, well being and increases energy. In musicians it is particularly effective with the communicative aspects of performance which are underpinned by confidence, and this we have shown even with novice abilities, i.e. singing in instrumentalists with no particular desire to sing. Slow wave training has impacted, though, on all aspects of performance including breath control and pitch. However, it may be fast wave training, SMR in particular, that has important applications in sports involving skilled visuomotor activities.

Overall, EEG-biofeedback shows promise in offering a means of optimising function that may have a sporting application. However, there is a dearth of controlled studies directly investigating this possibility, with further research clearly warranted [98].

7. Concluding remarks

EEG represents a useful methodological tool in understanding cortical processes that underlie performance in sporting and non-sporting domains. Although EEG lacks the spatial resolution of more expensive methods such as MEG or fMRI, it offers excellent temporal resolution and with advances in wireless hardware and equipment portability, allows a freedom of movement almost impossible to achieve with other neuroimaging technologies. Recording EEG during motion does present a number of problems with respect to obtaining 'clean' cerebral data. However, careful attention to proper methodological practices and developments in hardware and computational processing models offer a promising means of minimising, if perhaps not entirely eradicating, these problems.

Acknowledgements

The authors were in receipt of grants from the European PRESENCCIA project (IST-027731), the National Endowment of Science, Technology & Arts, U.K., and Brain Health, London.

References

[1] D.M. Ahrendt, Am. Fam. Physician 63 (2001) 913–922.
[2] D.T. Max, Wired for Victory. Retrieved 13th June 2008. Available from: <http://www.mensvogue.com/health/articles/2006/12/18/mindroom>.
[3] A.J. Rowan, E. Tolunsky, Primer of EEG, Elsevier, Philadelphia, USA, 2003.
[4] P.L. Nunez, R. Srinivasan, in: Electric Fields of the Brain: The Neurophysics of EEG, Oxford University Press, New York, 2006.
[5] G. Rippon, in: C. Senior, T. Russell, M.S. Gazzaniga (Eds.), Methods in Mind, MIT Press, MA, 2006, pp. 237–264.
[6] D.J. Vernon, Appl. Psychophysiol. Biofeedback 30 (2005) 347–364.
[7] R.D. Pascual-Marqui, C.M. Michel, D. Lehmann, Int. J. Psychophysiol. 18 (1994) 49–65.
[8] D.P. Burbank, J.G. Webster, Med. Biol. Eng. Comput. 16 (1978) 31–38.
[9] E. Seitsonen, A. Yli-Hankala, K. Korttila, Acta Anaesthesiol. Scand. 44 (2000) 1266–1270.
[10] A. Ravina, Presse Med. 60 (1952) 1575.
[11] E.W. Busse, A.J. Silverman, J. Am. Med. Assoc. 149 (1952) 1522–1525.
[12] M. Kaste, T. Kuurne, J. Vilkki, K. Katevuo, K. Sainio, H. Meurala, Lancet 2 (1982) 1186–1188.
[13] A.T. Tysvaer, Sports Med. 14 (1992) 200–213.
[14] A.T. Tysvaer, O.V. Storli, Am. J. Sports Med. 17 (1989) 573–578.
[15] A. Rutherford, R. Stephens, D. Potter, Neuropsychol. Rev. 13 (2003) 153–179.
[16] D.M. Landers, S.J. Petruzzello, W. Salazar, D.J. Crews, K.A. Kubitz, T.L. Gannon, M. Han, Med. Sci. Sports Exerc. 23 (1991) 123–129.
[17] W. Salazar, D.M. Landers, S.J. Petruzzello, M. Han, D.J. Crews, K.A. Kubitz, Res. Q. Exerc. Sport 61 (1990) 351–359.
[18] C. Babiloni, C. Del Percio, M. Iacoboni, F. Infarinato, R. Lizio, N. Marzano, G. Crespi, F. Dassù, M. Pirritano, M. Gallamini, F. Eusebi, J. Physiol. 586 (2008) 131–139.
[19] D.J. Crews, D.M. Landers, Med. Sci. Sports Exerc. 25 (1993) 116–126.
[20] M. Doppelmayr, T. Finkenzeller, P. Sauseng, Neuropsychologia 46 (2008) 1463–1467.
[21] C.H. Hillman, R.J. Apparies, C.M. Janelle, B.D. Hatfield, Biol. Psychol. 52 (2000) 71–83.
[22] N. Konttinen, H. Lyytinen, J. Sports Sci. 11 (1993) 257–266.
[23] L. Beyer, T. Weiss, E. Hansen, A. Wolf, A. Seidel, Int. J. Psychophysiol. 9 (1990) 75–80.
[24] T. Weiss, L. Beyer, E. Hansen, Int. J. Psychophysiol. 11 (1991) 203–205.
[25] G. Pfurtscheller, R. Leeb, C. Keinrath, D. Friedman, C. Neuper, C. Guger, M. Slater, Brain Res. 1071 (2006) 145–152.
[26] A. Dietrich, Methods 45 (2008) 319–324.
[27] J.A. Aarli, R. Vaernes, A.O. Brubakk, H. Nyland, H. Skeidsvoll, S. Tonjum, Acta Neurol. Scand. 71 (1985) 2–10.
[28] M.J. Halsey, Physiol. Rev. 62 (1982) 1341–1377.
[29] J.A. Kinney, R. Hammond, R. Gelfand, J. Clark, Electroencephalogr. Clin. Neurophysiol. 44 (1978) 157–171.
[30] B. Feddersen, H. Ausserer, P. Neupane, F. Thanbichler, A. Depaulis, R. Waanders, S. Noachtar, J. Neurol. 254 (2007) 359–363.
[31] T.P. Finnegan, P. Abraham, T.B. Docherty, Electroencephalogr. Clin. Neurophysiol. 60 (1985) 220–224.
[32] M. Beaumont, D. Batejat, C. Pierard, P. Van Beers, M. Philippe, D. Leger, G. Savourey, J.C. Jouanin, Sleep 30 (2007) 1527–1533.
[33] L. Nybo, B. Nielsen, J. Appl. Physiol. 91 (2001) 2017–2023.
[34] M.B. Pontifex, C.H. Hillman, Clin. Neurophysiol. 118 (2007) 570–580.
[35] Y. Nakamura, K. Nishimoto, M. Akamatu, M. Takahashi, A. Maruyama, Electromyogr. Clin. Neurophysiol. 39 (1999) 71–74.
[36] M.N. Magnie, S. Bermon, F. Martin, M. Madany-Lounis, G. Suisse, W. Muhammad, C. Dolisi, Psychophysiology 37 (2000) 369–377.
[37] C.H. Hillman, E.M. Snook, G.J. Jerome, Int. J. Psychophysiol. 48 (2003) 307–314.
[38] F. Grego, J.M. Vallier, M. Collardeau, S. Bermon, P. Ferrari, M. Candito, P. Bayer, M.N. Magnie, J. Brisswalter, Neurosci. Lett. 364 (2004) 76–80.
[39] S.R. Benbadis, in: T. Lee-Chiong (Ed.), Sleep: A Comprehensive Handbook, John Wiley & Sons, Colorado, USA, 2005.
[40] S.C. Schachter, D.L. Schomer, B.S. Chang, Atlas of Ambulatory EEG, Elsevier, MA, USA, 2005.
[41] M. Beaussart, J.D. Guiev, in: A. Redmond (Ed.), Handbook of Electroencephalography and Clinical Neurophysiology, Elsevier, Amsterdam, 1977, pp. 80–96 (11A).
[42] M. Saunders, in: D.W. Klass, D.D. Daly (Eds.), Current Practice of Clinical Electroencephalography, Raven Press, NY, 1979, pp. 37–68.
[43] J.L. Andreassi, Psychophysiology: Human Behavior and Physiological Response, Lawrence Erlbaum, NJ, 2000.
[44] L.P. Panych, J.A. Wada, M.P. Beddoes, Electroencephalogr. Clin. Neurophysiol. 72 (1989) 268–276.
[45] T.C. Ferree, P. Luu, G.S. Russell, D.M. Tucker, Clin. Neurophysiol. 112 (2001) 536–544.
[46] E. Freye, Cerebral Monitoring in the OR and ICU, Springer, New York, 2005.
[47] C. Falco, F. Sebastiano, L. Cacciola, F. Orabona, R. Ponticelli, P. Stirpe, G. Di Gennaro, Clin. Neurophysiol. 116 (2005) 1771–1773.
[48] C.D. Binnie, Clinical Neurophysiology, Butterworth, Heinemann, Boston, 1995.
[49] M. Matteucci, R. Carabalona, M. Casella, E. Di Fabrizio, F. Gramatica, M. Di Rienzo, E. Snidero, L. Gavioli, M. Sancrotti, Microelectron. Eng. 84 (2007) 1737–1740.
[50] M. Iwasaki, C. Kellinghaus, A.V. Alexopoulos, R.C. Burgess, A.N. Kumar, Y.H. Han, H.O. Luders, R.J. Leigh, Clin. Neurophysiol. 116 (2005) 878–885.
[51] J. Tieman, L.J. Peacock, K.J. Cureton, R.K. Dishman, Int. J. Neurosci. 106 (2001) 21–33.
[52] J.C. Smith, P.J. O'Connor, Biol. Psychol. 63 (2003) 293–310.
[53] F. Lecret, M. Pottier, Le Travail Humain 34 (1971) 51–68.
[54] J.A. Stern, J.A. Bynum, Aerosp. Med. 41 (1970) 300–305.
[55] J.A. Veltman, A.W.K. Gaillard, Biol. Psychol. 42 (1996) 323–342.
[56] M.D. Linderman, V. Gilja, G. Santhanam, A. Afshar, S. Ryu, T.H. Meng, K.V. Shenoy, Conf. Proc. IEEE Eng. Med. Biol. Soc. 1 (2006) 1212–1215.
[57] Mind Media, Flexible 32 Channel Monitoring & QEEG. Retrieved 15th June 2008, Available from: <http://www.mindmedia.nl/english/nexus32.php>.
[58] Interuniversity Microelectronics Centre, Wireless EEG system self-powered by body heat and light. Retrieved 15th June 2008. Available from: <http://www.sciencedaily.com/releases/2008/04/080412172006.htm>.
[59] M.J. Griffiths, P. Grainger, M.V. Cox, A.W. Preece (2005) in "3rd IEE International Seminar on Medical, Applications of Signal Processing".
[60] A. Bell, T. Sejnowski, Neural. Comput. 7 (1995) 1129–1159.
[61] T. Kochiyama, T. Morita, T. Okada, Y. Yonekura, M. Matsumura, N. Sadato, Neuroimage 25 (2005) 802–814.
[62] J.V. Stone, Trends Cogn. Sci. 6 (2002).
[63] C.G. Puntonet, A. Prieto, Independent Component Analysis and Blind Signal Separation, Springer, New York, 2004.
[64] A. Delorme, T. Sejnowski, S. Makeig, Neuroimage 34 (2007) 1443–1449.
[65] J. Onton, M. Westerfield, J. Townsend, S. Makeig, Neurosci. Biobehav. Rev. 30 (2006) 808–822.
[66] M. Crespo-Garcia, M. Atienza, J.L. Cantero, Ann. Biomed. Eng. 36 (2008) 467–475.
[67] R. Romo-Vazquez, R. Ranta, V. Louis-Dorr, D. Maquin, Conf. Proc. IEEE Eng. Med. Biol. Soc. (2007) 5445–5448.
[68] W. Zhou, J. Zhou, H. Zhao, L. Ju, Conf. Proc. IEEE Eng. Med. Biol. Soc. 6 (2005) 6017–6020.
[69] W. Nakamura, K. Anami, T. Mori, O. Saitoh, A. Cichocki, S. Amari, IEEE Trans. Biomed. Eng. 53 (2006) 1294–1308.
[70] A. Delorme, S. Makeig, J. Neurosci. Methods 134 (2004) 9–21.
[71] J. Onton, A. Delorme, S. Makeig, Neuroimage 27 (2005) 341–356.
[72] C. Hesse, C. James, IEEE Trans. Biomed. Eng. 53 (2006) 2525–2534.
[73] N. Hironaga, A.A. Ioannides, Neuroimage 34 (2007) 1519–1534.
[74] N.P. Castellanos, V.A. Makarov, J. Neurosci. Methods 158 (2006) 300–312.
[75] S. Boudet, L. Peyrodie, P. Gallois, C. Vasseur, Conf. Proc. IEEE Eng. Med. Biol. Soc. 1 (2006) 5719–5722.
[76] P. LeVan, E. Urrestarazu, J. Gotman, Clin. Neurophysiol. 117 (2006) 912–927.
[77] F. Shayegh, A. Erfanian, Conference Proceedings: Annual International Conference of the IEEE Engineering in Medicine and Biology Society. IEEE Engineering in Medicine and Biology Society. Conference 1 (2006) 5269–5272.
[78] N. Mourad, J.P. Reilly, H. De Bruin, G. Hasey, D. MacCrimmon, ICASSP, IEEE International Conference on Acoustics, Speech and Signal Processing—Proceedings, 1, Art. No. 4217099 (2007) I393–I396.
[79] E.I. Bird, Int. J. Sports Psychol. 18 (1987) 9–18.
[80] S.J. Radlo, G.M. Steinberg, R.M. Singer, D.A. Barba, A. Melinkov, Int. J. Sports Psychol. 33 (2002) 205–217.

[81] T. Steffert, A. Steed, J. Leach, T. Thompson, J. Gruzelier, Revista Espanola de Neuropsicologia 10 (2008) 71–77.
[82] A.J. Haufler, T.W. Spalding, D.L. Santa Maria, B.D. Hatfield, Biol. Psychol. 53 (2000) 131–160.
[83] M. Arns, M. Kleinnijenhuis, K. Fallahpour, R. Breteler, J. Neurother. (in press)..
[84] D. Vernon, T. Egner, N. Cooper, T. Compton, C. Neilands, A. Sheri, J. Gruzelier, Int. J. Psychophysiol. 47 (2003) 75–85.
[85] T. Egner, J.H. Gruzelier, Neuroreport 14 (2003) 1221–1224.
[86] T. Egner, J.H. Gruzelier, Neuroreport 12 (2001) 4155–4159.
[87] T. Thompson, T. Steffert, E. Redding, J. Gruzelier, Revista Espanola de Neuropsicologia 10 (2008) 59–62.
[88] M.B. Sterman, L.R. Macdonald, R.K. Stone, Epilepsia 15 (1974) 395–416.
[89] J.D. Kropotov, V.A. Grin-Yatsenko, V.A. Ponomarev, L.S. Chutko, E.A. Yakovenko, I.S. Nildshena, Int. J. Psychophysiol. 55 (2005) 23–24.
[90] M. Foks, Educ.Child Psychol. 22 (2005) 67–77.
[91] W. Klimesch, Brain Res. Brain Res. Rev. 29 (1999) 169–195.
[92] D. Vernon, A. Frick, J. Gruzelier, J. Neurother. 8 (2004) 53–82.
[93] T. Egner, J.H. Gruzelier, Clin. Neurophysiol. 115 (2004) 131–139.
[94] J. Gruzelier, T. Egner, D. Vernon, in: C. Neuper, W. Klimesch (Eds.), Event-Related Dynamics of Brain Oscillations, 159, Elsevier Science, Amsterdam, 2006, pp. 421–431.
[95] T. Ros, P. Bloom, L. Benjamin, M. Moselely, L. Parkinson, J. Gruzelier, Revista Espanola Neuropsicol. 10 (2008) 97–101.
[96] T. Boynton, J. Neurother. 5 (2001) 5–18.
[97] J. Raymond, I. Sajid, L.A. Parkinson, J.H. Gruzelier, Appl. Psychophysiol. Biofeedback 30 (2005) 65–73.
[98] D.C. Hammond, J. Am. Board Sport Psychol. (2007) 1–9.