Brownian motion. As discussed above, the kinetic energy associated with this motion is proportional to temperature (i.e., 3/2 kT). Accordingly, to maintain a trap, the average magnetic potential energy of the generated field must be sufficiently greater than the sample's thermal energy.

[0142] In view of the foregoing, once a sample is initially trapped based on a pulsed magnetic field, the sample may remain trapped in the pulsed magnetic field as long as the magnetic field is not off for a period of time that allows significant diffusion of the sample away from the "trapping area" above a given microcoil. An upper limit for the field off-time τ_{off} is given approximately by $\tau_{off}<d^2/D$, where d is the diameter of the microcoil and D is the diffusion constant of the sample (from the definition of D, for a given time t, a particle travels an average distance $d=(Dt)^{1/2}$). The diffusion constant D of a sample (in meters^2 per second) is given generally by

$$D=kT/3\pi\eta a \quad (3)$$

where η is the viscosity of the liquid (in kg/m·s) and a is the diameter of the sample. In the exemplary scenario under consideration, the viscosity η of water is approximately 10^{-3} kg/m·s and the diameter of the Dynabead sample is 5 μm; accordingly, assuming a temperature T of approximately 300 K (i.e., room temperature), the diffusion constant D for the Dynabead sample in water is approximately 8.5×10^{-14} m^2/s. If a microcoil diameter of 20 μm is assumed, τ_{off} should be less than approximately 5000 seconds. From a practical standpoint, the foregoing example illustrates that multiplexing current to the microcoils at a rate of 10,000 Hz or higher (i.e., $\tau_{off}<10^{-4}$ seconds) permits practically no appreciable diffusion of the sample due to Brownian motion; with an off-time $\tau_{off}<10^{-4}$ seconds, the 5 μm Dynabead diffuses approximately only 3 nanometers.

[0143] In general, it should be appreciated that the configuration of current sources and microcoils illustrated in **FIGS. 13-16** and the multiplexing technique described above are provided as an exemplary implementation, and that other configurations according to the present disclosure are possible. For example, in alternative configurations, the array **200**B may be subdivided into greater or fewer subdivisions (e.g., four microcoil cells per subdivision instead of sixteen), wherein a variable current having a predetermined number of different current levels for each subdivision is provided by one current source dedicated to the subdivision. Alternatively, in another implementation, only one such current source may provide current to all the microcoil cells of the array **200**B in a sequential time-shared (e.g., multiplexed) manner. In yet another configuration, each microcoil cell may be equipped with its own variable current source, such that there is no need to multiplex one current source amongst multiple microcoils. In general, any implementation that makes use of a current-sharing scheme by using one current source to provide current to multiple microcoils reduces DC power dissipation from the system.

[0144] It should also be appreciated that while the exemplary concepts discussed above in connection with **FIGS. 13-16** focus on microcoils driven by current sources, alternative implementations of field-generating arrays for sample manipulation based on the general switching and multiplexing architecture outlined in **FIGS. 13-16** may be based on electric-field generation and dielectrophoresis principles using microcoils or microposts driven by voltage sources.

[0145] For example, first consider the microcoil array **200**B of **FIG. 13** and associated control components, with a substitution of one or more variable (or fixed) voltage sources for the variable current sources. In one such exemplary implementation, respective microcoil cells **250** of the array are selected/enabled in the same manner described above in connection with **FIGS. 13-15**. However, rather than passing a variable current though a selected/enabled microcoil, a variable voltage (having a selectable polarity based on the direction signal **472**) may be connected to the selected/enabled microcoil to generate a corresponding electric field from the microcoil (e.g., a variable voltage source may replace the variable current source **420-1** shown in **FIG. 14**, and upon selecting/enabling a given microcoil and the microcoil polarity via the signal **472**, a variable voltage would essentially be placed in series with the power supply voltage Vdd across the microcoil). One or more variable voltage sources of such an alternative implementation may be realized by any number of conventional configurations (e.g., a digital-to-analog converter) suitable for various integrated circuit fabrication processes.

[0146] In another example based on electric field generation, the microcoil array **200**B of **FIG. 13** may be substituted by an appropriately-sized micropost array similar to that discussed above in connection with FIGS. **5**(*a*) and (*b*), and again the variable current sources would be substituted by one or more variable (or fixed) voltage sources. In yet another example, the microcoil array **200**B of **FIG. 13** may be employed with both variable current sources and variable voltage sources to provide a subsystem capable of sample manipulation based on both electric and magnetic fields. These and other types of electric field-based or electric/magnetic field-based implementations may be employed for a variety of applications relating to manipulation, sensing and imaging systems that integrate microelectronics and microfluidics.

[0147] III. Sample Detection, Imaging and Characterization

[0148] As discussed above in Section I, with reference again to **FIG. 1**, the field control components **400** of a semiconductor-based/microfluidic hybrid system additionally may include radio frequency (RF) and other detection components **480**, coupled between the field-generating components **200** and the one or more processors **600**, for facilitating one or more of detection, imaging and characterization of samples contained in the microfluidic system **300**, according to various embodiments of the present disclosure. In different aspects, the RF/detection components **480** are configured to facilitate both the generation of electromagnetic fields from the field-generating components based on relatively high frequency (e.g., RF, microwave) electric signals (voltages or currents), as well as the measurement of signals indicating some type of interaction between the generated RF fields and one or more samples of interest.

[0149] In general, as is well known based on Maxwell's Equations, an RF field is capable of interacting with virtually any particle (biological or otherwise) that conducts electricity at the RF signal frequency, or is polarizable electrically or magnetically. Accordingly, in various embodiments of the present disclosure, the interaction between RF electric and/or magnetic fields and samples of interest may be exploited